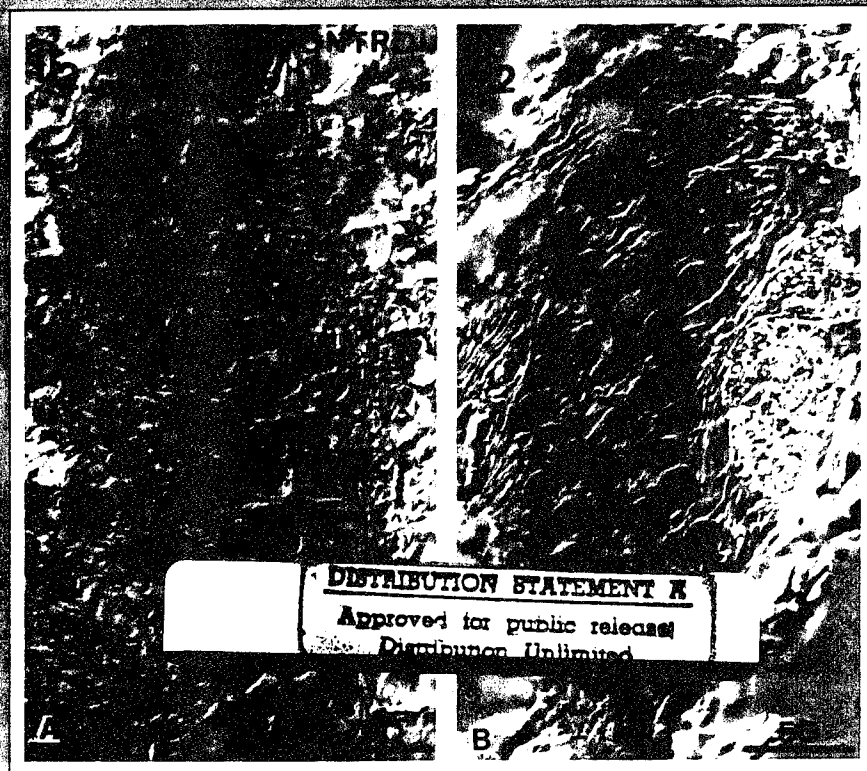


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Scientific Basis of Noise-Induced Hearing Loss

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Scientific Basis of Noise-Induced Hearing Loss

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Preface

The twentieth century has witnessed the continual increase and spread of industrialization throughout the world. Accompanying this development has been a growing increase in noise levels and the number of individuals exposed to hazardous levels of noise often in combination with various other physical/chemical agents. The levels of noise now common in many industrial/military environments place an excessive burden on the auditory system. Recognizing the problems engendered by excessive noise exposure, individuals such as Temkin and others in the early decades of this century made remarkable progress in understanding noise-induced hearing loss (NIHL). However, it was not until governments finally acknowledged the epidemic proportions of NIHL and the concomitant impact that hearing loss has on productivity and the quality of life, that they established regulatory and funding agencies to reduce exposure levels and to provide a mechanism to support basic and applied research on the biological effects of noise. As a result, our knowledge of NIHL has increased exponentially in the post-World War II period. New and unanticipated phenomena were discovered and continue to be studied, for example, asymptotic threshold shift; the modulation of threshold shift by intermittent exposure paradigms or a priming preexposure; and genetic influences on NIHL; to name a few. New diagnostic approaches evolved from various psychoacoustic and physiological studies (brain stem audiometry, otoacoustic emissions) and new noise measurement technology and protective systems (active noise reduction) were developed. Now in the final decade of the century, there is great excitement and optimism at the prospect of inducing the regeneration of sensory cells destroyed by noise. These and other related topics are the subject matter of this volume, the fifth in a series that began in 1975. This collection of five volumes of symposium proceedings provides a comprehensive overview and a continuing updating of our understanding of the biological effects of noise and strategies for controlling the adverse effects. The volume contains the text of 36 presented papers, which are grouped into the four sections described below.

Biological Basis of Noise-Induced Hearing Loss (NIHL)

This section focuses on the biological/physiological basic science aspects of NIHL. New findings are presented on sensory cell regeneration, the mechanics of the noise-damaged organ of Corti, genetic contributions to NIHL, biochemical changes following trauma, and the behavior of otoacoustic emissions.

Experimental Studies of Noise-Induced Hearing Loss

New laboratory-based (animal model) noise exposure studies are presented that explore issues from aging to noise effects in exotic environments, for example,

PREFACE

underwater NIHL and the fetal response to noise. The modulation of threshold shifts by priming exposures and interrupted exposure paradigms offer new insights into the cochlear response to noise, and new approaches to noise measurement hold the prospect of making better estimates of noise dose and predicting eventual permanent threshold shifts from noise exposures.

Auditory Performance Changes With Noise-Induced Hearing Loss

Understanding the changes in psychoacoustic performance in individuals with NIHL provides a better perspective on information processing in the normal auditory system, the development of new diagnostic procedures, and understanding the implications of abnormal signal processing for worker safety.

Human Studies of Noise-Induced Hearing Loss

This section focuses primarily on human experimental and demographic studies with an emphasis on hearing protector performance in various noise environments. The section also contains chapters that focus on the sociological aspects of NIHL, such as compensation and hearing conservation strategies.

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The Vth International Symposium on the Effects of Noise on Hearing was held in Göteborg, Sweden, on May 12–14, 1994. The organizers, Alf Axelsson, Hans Borchgrevink, Roger P. Hamernik, Per-Anders Hellström, Donald Henderson, and Richard J. Salvi, wish to thank the contributors for their substantial support and for their contribution to the success of this conference.

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Section /

Biological Basis of Noise-Induced Hearing Loss

Chapter 1

Sensory Cell Regeneration and Functional Recovery: A Review

Andrew Forge

Hair cells in the lateral line systems of fish and amphibians are continuously produced as the animals grow, and lost hair cells can be replaced after injury even in the most dramatic of circumstances.^{1,2} Likewise, in the inner ears of these lower vertebrates, constant hair cell production results in continuously increasing numbers³⁻⁵ and enables recovery after trauma-induced damage.^{6,7} In contrast, it was believed that in warm-blooded vertebrates, birds and mammals, in which there is no postembryonic growth of the inner ear epithelia, generation of hair cells occurs only during embryonic life.⁸ Thus, any hair cells subsequently lost as a natural consequence of aging or through exposure to noise, ototoxic agents, or infection are irreplaceable. This leads to permanent functional deficits. Consistent with this notion is that despite investigations of numerous aspects of mammalian cochlear pathology over many years, there is no evidence for the spontaneous replacement of lost hair cells in the mature mammalian organ of Corti; indeed, examination of the cochleae from humans with no known auditory dysfunction suggests there is a continuous decline in hair cell number throughout life in normal individuals.⁹

However, it is now clear that the basilar papilla (auditory sensory epithelium) of birds has retained the capacity to replace those hair cells lost, through injury caused by noise trauma or ototoxic drugs, with functional new hair cells. Three kinds of evidence led to this discovery. First, morphological observations showed recovery of hair cell num-

bers after losses induced by acoustic overstimulation¹⁰ or ototoxic aminoglycoside damage,¹¹ and cells exhibiting all stages of normal maturation of hair cell stereociliary bundles during the postnoise trauma period prior to the reestablishment of a normal appearing epithelium.¹⁰ Second, studies using tritiated thymidine, which is incorporated into DNA during its synthesis prior to cell division and thereby provides a radioactive label to the nuclei of proliferating cells and their progeny. Administration of tritiated thymidine to birds after traumatizing noise^{12,13} or ototoxic doses of aminoglycoside¹⁴ resulted in the appearance of hair cells with radioactively labeled nuclei. This demonstrated that hair cells repopulating the epithelium were newly produced. Third, studies demonstrating recovery to near normal auditory thresholds after initial deficits induced by either noise¹⁵ or aminoglycoside¹⁶ suggested that sensory function returned.

These findings indicate that following trauma-induced injury some population of cells that normally do not proliferate, is stimulated to enter the cycle of cell division and undergo mitosis. One or both of the immediate or succeeding progeny then differentiate into new hair cells. These regenerated hair cells could provide a basis for functional recovery. As discussed at the previous meeting in this series¹⁷ and in other reviews,¹⁸⁻²⁰ the discovery of hair cell regeneration in birds suggests that a distinction between "lower" and "higher" vertebrates with respect to their abilities to produce hair cells postembryon-

ically is no longer valid and has raised the prospect of replacement of hair cells in the damaged mammalian organ of Corti.

General Pattern of Hair Cell Loss and Recovery in the Avian Basilar Papilla

Morphology of Basilar Papilla

Detailed accounts of the structure of the avian basilar papilla have been published.^{20–22} Briefly, the basilar papilla consists of a curved sheet of cells that is quite narrow at the proximal (basal) end, where high frequency sounds are detected, progressively widening toward the distal (apical) low frequency end. Nerves enter the epithelium along the outside (neural) edge of the curve. Like other hair cell containing sensory epithelia, the surface of the basilar papilla appears as a regular mosaic formed of hair cells and supporting cells. The hair cells have an approximately hexagonal apical surface and each one is separated from its neighbor by intervening supporting cells. There are systematic variations in the height and number of stereocilia across and along the papilla²³ and the hair bundles on each hair cell are oriented with the longest stereocilia and kinocilium directed toward the abneural edge of the papilla. Two types of hair cell are recognized: short hair cells, which are located over the basilar membrane on the abneural side of the epithelium and have an almost exclusively efferent innervation; and tall hair cells, which are situated over a cartilaginous shelf on the neural side of the papilla and have a predominantly afferent innervation. The papilla is covered with the fibrous tectorial membrane. Thus, there is a general similarity between the basilar papilla and the organ of Corti, for example tall and short hair cells can be considered to correspond to inner and outer hair cells respectively, but supporting cells in the basilar papilla are structurally different from those of the organ of Corti. They do not possess the complex cytoskeletal framework that is seen in organ of Corti supporting cells and each one closely apposes and surrounds the bodies of the hair cells, unlike the organ of Corti where specialization of the supporting

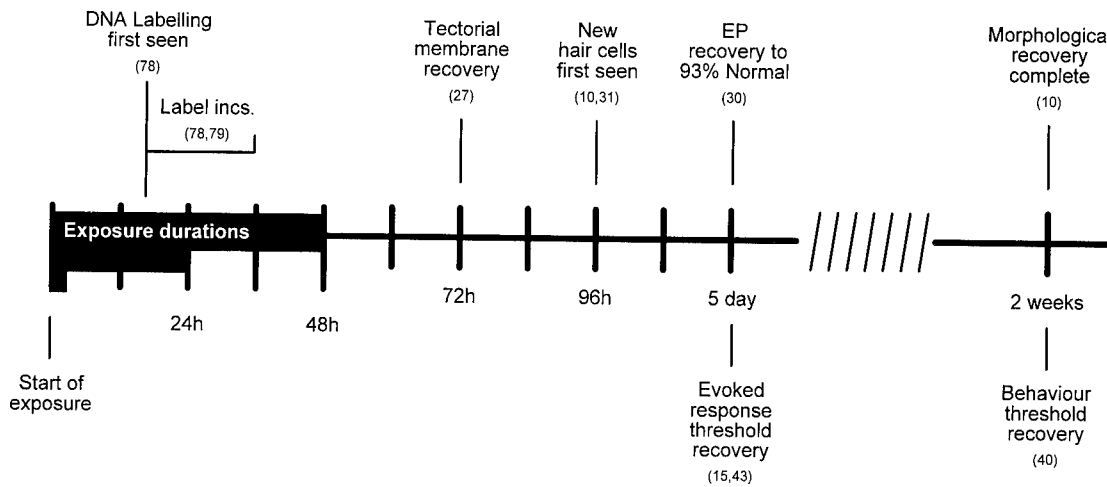
cells creates large extracellular spaces surrounding the bodies of the outer hair cells. In this respect, the avian basilar papilla more closely resembles vestibular sensory epithelia and the auditory epithelia of lower vertebrates.

Injury and Recovery With Noise Exposure and Ototoxic Drug Damage

Acoustic overstimulation causes damage to the basilar papilla in a region that is related to the frequency of the damaging sound. An excellent, brief account of details of the effects of varying exposure conditions can be found in a recent review by Cotanche et al.²⁰ The damage is initiated at the abneural edge of the lesion site, that is, in the region of short hair cells, and spreads during prolonged exposure toward the neural edge and in the basal and apical directions to create a semicircular “patch” lesion which predominantly affects only the short hair cells.²⁴ Within the lesion, loss of short hair cells occurs, but about 65–70% of them remain, although many of these show abnormalities indicating injury.^{10,25,26} Noise trauma also causes damage to the tectorial membrane which is completely disrupted over the site of the patch lesion in the epithelium.^{27,28} In addition, the tegmentum vasculosum, the ion-transporting epithelium that maintains a positive potential within endolymph, is affected by noise resulting in a decline in endocochlear potential (EP).^{29,30}

Cells with immature hair bundles can be identified within the lesioned area by 4 days from the start of exposure^{10,31} and a normal appearing epithelium is present by 2 weeks postexposure^{10,25} (Figure 1-1). There is some evidence, however, that the number of hair cells in the recovered epithelium is less than normal.^{13,26} This suggests that not every lost hair cell is replaced, and that the recovery in the morphological appearance of the epithelium derives, in part, from shape changes in surviving hair cells and supporting cells.

Aminoglycoside antibiotics, in contrast to noise trauma, cause the loss of both short and tall hair cells, but do not cause obvious disruption of the tectorial membrane.^{32–35} The dam-



INTENSITIES : 115-125dB SPL (Above 125dB; complete epithelial destruction may occur)

DURATION : 4h - 48h (Area of damage progressively increases up to ca. 40h., then no apparent further increase)

LOCATION OF DAMAGE : Patch at position related to frequency of damaging noise.

NATURE OF DAMAGE : Short hair cells affected (maybe ca. 65% retained).
 Tall hair cells (afferent innervation) appear undamaged.
 Tectorial membrane disrupted at same site as hair cell injury.

Figure 1-1 Sequence of events in hair cell regeneration and functional recovery after noise trauma in the avian basilar papilla.

age is initiated at the basal end of the papilla and spreads progressively toward the apex (Figure 1-2). Studies of the effects of aminoglycosides on the basilar papilla have used either gentamicin^{11,16,32,33} or kanamycin³⁴⁻³⁷ and there is some difference in the extent and progression of damage caused by these agents (Figure 1-2). Gentamicin appears to be much more potent with hair cell loss progressing over a period of about 4-5 weeks after the end of treatment, to affect about 60% of the papilla. The damage caused by kanamycin does not progress for more than about 1 week after the end of 10 days of treatment and results in total hair cell loss at the basal end for ca. 40% of papilla length, and partial damage over a further ca. 20% of its length.

With gentamicin, near complete recovery in hair cell number is not apparent until 10-20 weeks,^{32,33} but with kanamycin recovery is nearly complete by about 14 days from the end of treatment.^{34,35} Immature hair cells are first

observed in the basal region of the papilla.^{32,35} Following kanamycin treatment there appears to be a progressive maturation in the basal to apical direction, paralleling the progression of hair cell loss.^{34,35} After gentamicin treatment the apical end of the lesion seems to repair itself before the basal region, suggesting repair in the opposite direction to that of hair cell loss.³² However, because immature hair cells are first observed in the basal end while damage is still progressing apically (Figure 1-2), it may be that the high potency and prolonged effect of gentamicin results in continuing damage to the new hair cells in the basal region as they mature. Thus, several "rounds" of regeneration may occur. Immature hair cells in the basilar papilla are also sensitive to aminoglycosides and this can result in developmental abnormalities of the apical structures of the hair cells.³⁸ In the most basal region of recovered papillae, the hair bundles, which in normal tissue are oriented uniformly

Kanamycin

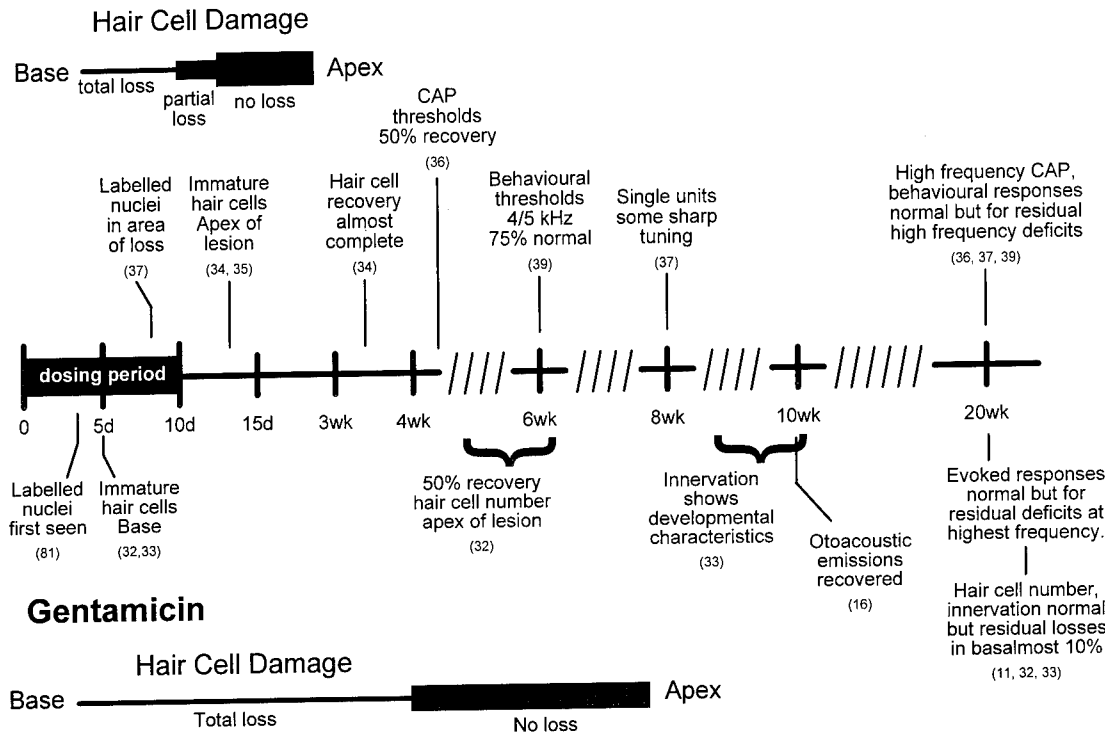


Figure 1-2 Sequence of events in hair cell regeneration and functional recovery after ototoxic aminoglycoside-induced hair cell loss in the avian basilar papilla. Results from studies with kanamycin are represented above the time line, and studies with gentamicin below the line. The schematics depicting damage from each represent the time periods over which damage occurs measured against the time line, and the extent of damage that ultimately may occur along the length of the basilar papilla for each aminoglycoside.

on different cells, are often misaligned suggesting developmental impairment.³⁹ Furthermore, from studies of sound-evoked behavioral responses, which can be tested in the same individual over prolonged periods, it is apparent that after recovery from the initial damage caused by aminoglycoside or by noise, hair cell regeneration and functional recovery can take place again after a second similar insult.^{39,40}

Innervation of Regenerated Hair Cells and Functional Recovery in Basilar Papilla

For the regenerated sensory cells to contribute to the functional recovery that is observed re-

quires that they become innervated. That they do so after noise trauma has been demonstrated in elegantly simple fashion.⁴¹ Serial sections were obtained from the basilar papillae of quail that had been administered tritiated thymidine after noise exposure. Light microscopy of sections prepared for autoradiography enabled identification of regenerated hair cells by the presence of radioactivity in their nuclei. Electron microscopy of adjacent thin sections showed mature synaptic connections on these same cells. Following gentamicin damage, it has been observed that neural endings are associated with cells with characteristics of immature hair cells.³³ Furthermore, examination at various recovery times of short hair cells at locations where the

drug initially caused extensive hair cell loss, and thus were almost certainly regenerated hair cells, showed progressive alterations in their innervation that mimicked those seen during normal embryonic development of the basilar papilla. An initial afferent innervation subsequently degenerated and was replaced by efferent endings. Although the hair cells examined here were not positively identified as regenerates, these results indicate newly developing hair cells acquiring appropriate innervation.

Thus, these new hair cells should potentially be able to convey sensory information to the auditory neural pathway. However, there is evidence from quail that after noise exposure, the number of neural ganglion cells continues to decline even after hair cells have been replaced in the epithelium.⁴² This might be expected to limit the extent and nature of functional recovery. As yet, the somewhat difficult experiment of determining the neural outputs from defined regenerated cells has not been performed, although studies of single unit responses after ototoxic damage³⁷ come close to this. Rather, the progression of recovery of different parameters of auditory function in comparison with each other and with the progression of epithelial recovery has been examined (Figures 1-1, 1-2).

Following noise trauma, initial studies of sound-evoked potentials in the nucleus magnocellularis (cochlear nucleus) showed a threshold shift of approximately 60 dB after the end of a 48 hour exposure recovered to ca. 15 dB of shift by 2 weeks,¹⁵ roughly following the time course of recovery in hair cell number. However, more intensive physiological assessment showed a remarkable degree of recovery both in thresholds and in tuning properties within 3 days of the end of exposure,^{15,43} very soon after immature hair cells are first seen and well before full morphological recovery of the epithelium (Figure 1-1). Likewise behavioral studies in quail, which enable an assessment of the ability of the animal to "hear" and use that information, have shown significant recovery of the behavioral thresholds at times preceding that at which complete recovery of hair cells occurs.⁴⁰ The recovery of behavioral

thresholds implies a "useful" sensory output from the epithelium and an intact auditory pathway. This correlates with the maintenance of tonotopicity in the nucleus magnocellularis after noise exposure.⁴⁴

The recovery of evoked potential and behavioral thresholds before restoration of normal epithelial morphology suggests factors other than regenerated hair cells may be involved in functional recovery. The disrupted tectorial membrane is repaired quite rapidly by secretions from the supporting cells.^{27,28} (Supporting cells are not damaged by the sound exposure.) Within 24 hours of the end of a 48 hour exposure, a fibrillar, honeycomb-like lattice is created with elements surrounding each surviving short hair cell and contacting the tallest stereocilia of each one across the patch lesion.²⁷ Although the tectorial membrane does not fully regrow even after prolonged postexposure periods, it has been argued that this early repair, and perhaps also the retention of significant numbers of short hair cells, is instrumental in renewal of appropriate stimulation of the tall hair cells in response to basilar membrane movement⁴⁵; the tall hair cells remain largely undamaged by the noise exposure in the first place and almost all of the basilar papilla's afferent innervation derives from them. An alternative explanation is that function returns with recovery of the tegmentum vasculosum and EP.³⁰ It has been suggested that initial functional deficits may derive from alterations of the ionic environment around the tall hair cells occasioned by loss of the short hair cells and concomitant leakage of endolymph through the apical surface of the epithelium. With repair of the epithelium and recovery of EP, the activity of tall hair cells is restored. Recovery of EP may begin within 24 hours of the end of noise exposure.^{29,30} However, hair cell loss is accompanied by repair of the lesion by supporting cells in a manner that appears to prevent the formation of obvious lesions through the permeability barriers at the apical surface of the basilar papilla⁴⁶ (see later). If this is so, there may in fact be relatively little disturbance of the ionic environment within the *corpus* of the papilla, so that tall hair cell

functioning would be only minimally disturbed and EP could be rapidly reestablished.

Both of these explanations for functional recovery after noise trauma suggest that the regenerated short hair cells contribute relatively little to the return of function. This begs the question of what the role of the short hair cells might be and why energy should be expended in their regeneration. More detailed studies of functional recovery after noise trauma may well provide answers to this question.

In contrast, because aminoglycoside-induced injury does not obviously affect the tectorial membrane, and the tall hair cells as well as the short hair cells are lost, functional recovery is more likely associated with the production of new hair cells. With the prolonged time course of recovery that follows gentamicin-induced hair cell loss, it has been possible to demonstrate recovery of otoacoustic emissions at about the time that fully mature short hair cells repopulate the epithelium, but prior to the establishment of mature innervation.^{16,17} Sound-evoked potential thresholds subsequently return close to normal, at times when innervation patterns also appear normal.^{16,32} Following kanamycin treatment of chicks, compound action potential (CAP) and single unit thresholds show significant recovery within 7–14 days of the reestablishment of normal appearing epithelium³⁶ and are succeeded by recovery in tuning,³⁷ although this takes some time to fully reappear. Behavioral thresholds, examined in studies performed with starlings³⁹ also show substantial recovery around the time or even preceding the recovery of tuned single unit responses in chicks. Thus, auditory function returns progressively after hair cell regeneration. Moreover, because behavioral thresholds recover, the birds are able to “hear.” Further definition of the quality of the sound perceived with regenerated hair cells could be undertaken with behavioral studies using starlings because these are song birds.³⁹ One feature observed in all studies of recovery after aminoglycoside injury, however, is a persistent functional deficit at the highest frequencies (ca. 4 kHz in studies with chicks,^{36,37} 7 kHz

with starlings³⁹). This is associated with persistent hair cell loss and abnormalities in hair bundle orientation at the basal end of the papilla.³⁹ There may therefore be some limitation of the extent to which regeneration can occur.

Hair Cell Regeneration in Avian Vestibular System

Postembryonic production of hair cells occurs in birds, not only after trauma-induced losses in the basilar papilla, but also in the vestibular sensory epithelia. As in other vertebrates, including mammals, the vestibular sensory epithelia in birds are present in the three cristae, one in each of the three semicircular canals, and in the maculae of the utricle and saccule. In the avian sensory epithelia, there are two types of hair cell. The Type 1 cells are flask-shaped and possess a single afferent nerve calyx that encloses the cell body. The Type 2 hair cells, which are considered the less differentiated form, are cylindrical and have afferent and efferent bouton-like endings synapsing at the base. The bodies of the supporting cells surrounding the nuclei lie below the level of the hair cells and send thin processes between the hair cells to the apical surface of the epithelium where they intervene between adjacent hair cells. This structural organization of the avian vestibular system is almost identical to that of mammals and very similar to that of lower vertebrates. Although once it was thought that fish and amphibia possessed only one type of vestibular hair cell, similar to the Type 2 hair cell of higher vertebrates, recent studies of fish⁴⁷ have shown two types of hair cell, one of which is similar in many respects to avian and mammalian Type 1 hair cells. In fish and amphibians, constant production of hair cells occurs throughout life and hair cells lost as a consequence of ototoxic aminoglycoside damage are replaced.^{6,7}

Constant production of hair cells also occurs in the avian vestibular system. Proliferation of cells in the undamaged sensory epithelia of adult budgerigars has been demonstrated.⁴⁸ More extensive studies of chicks⁴⁹ then noted that in normal animals, the nuclei both of sup-

porting cells and of Type 2 hair cells showed radioactive labeling after tritiated thymidine injections, or immunocytochemical labeling after injections of bromodeoxyuridine (BrdU), another analogue of one of the DNA bases that is incorporated during DNA synthesis and for which there are specific antibodies. The newly produced cells appeared throughout the epithelium and labeled hair cell nuclei were usually present above labeled supporting cell nuclei, suggesting that mitotic division of supporting cells gave rise to new hair cells. The production of new hair cells throughout life ought to lead to increasing numbers of cells and growth of the epithelium unless there is compensatory cell loss, the situation that obtains in other proliferating tissues to regulate cell numbers. The avian vestibular epithelia do not appear to grow throughout life and degenerating cells have been observed in the normal epithelia⁴⁸ (and in the normal mammalian vestibular system⁵⁰), so it is likely that there is a continuous turnover of cells with the loss of cells stimulating the production of new ones. Some evidence has also been presented to suggest that there may be some cell turnover in the undamaged basilar papilla as well.⁵¹ The rate of proliferation was, however, extremely low. The proliferation was almost entirely limited to the apical third of the papilla and the labeled cells within the epithelium were almost all identified as supporting cells.

In these circumstances, it is not really surprising that regeneration of hair cells has been observed in the avian vestibular system after aminoglycoside-induced injury.⁵² After repeated systemic treatment with streptomycin, hair cell loss occurred progressively over a period of 3 weeks. Almost complete recovery of hair cell numbers was apparent by 8–9 weeks after the end of treatment. Tritiated thymidine labeling showed labeled nuclei 1 day after the end of 7 days of drug treatment and labeled hair cells at 20 days posttreatment. Predominantly, Type 1 hair cells were lost, but during recovery, cells with morphological characteristics similar to Type 2 hair cells appeared. At later recovery times, morphologically definable Type 1 cells were also

present and the total number of hair cells was greater than that in normal tissue, indicating some over production. The sequential appearance of the two hair cell types could mean Type 1 cells arise from continued differentiation of Type 2's, or that the two hair cell types develop independently but that their immature forms initially show similar morphological characteristics that resemble mature Type 2 hair cells before further differentiation of the Type 1 hair cells results in the acquisition of their distinguishing features.

Recent studies of CAP responses from the vestibular nerve,⁵³ and of the vestibulo-ocular reflexes (VOR)⁵⁴ in chicks which had received streptomycin, have also suggested functional recovery of the vestibular sensory epithelia can occur. After initial drug-induced deficits, restoration of CAP thresholds and recovery of VOR occurred over time courses corresponding to that of epithelial repair, suggesting that the new hair cells contribute to functional recovery.

Regeneration in Mammalian Inner Ear

The structural characteristics of the fish, amphibian, and avian vestibular sensory epithelia are almost identical to those of mammals. It is, therefore perhaps not surprising, although it was unexpected, to find characteristics of hair cell regeneration after gentamicin-induced losses in the vestibular sensory epithelia of guinea pigs.^{55,56} After either repeated systemic dosing or a single, topical application of the drug to the middle ear cavity, loss of hair cells from the central regions of the epithelia was apparent by 1 week posttreatment. By 4 weeks, in those areas where extensive hair cell loss had occurred earlier, there were cells with characteristics typical of maturing hair bundles at all stages of development. There was also a recovery in cell numbers as assessed from counts of hair bundles using scanning electron microscopy (SEM): at 1 week posttreatment, there was about one-third the normal number; at 4 weeks about half; and by 12 weeks about two-thirds. Thus, although there was restitution of numbers, it was incomplete at 3 months posttreatment. It was

also apparent that many of the hair bundles at 12 weeks were not fully mature, often being noticeably shorter than their counterparts in age-matched controls. Thin sectioning showed loss of Type 1 hair cells; but at 4 weeks post-treatment, many immature hair cells, most of which showed morphological characteristics resembling Type 2's, were present. Counts of hair cells at 4 and 12 weeks showed a recovery in the number of hair cells present similar to that observed by SEM, and confirmed significantly greater numbers of Type 2-like hair cells than were present in age-matched controls at the same locations. These findings suggested that lost Type 1 cells were replaced by newly developing hair cells. There were also more morphologically distinguishable Type 1 cells present at 12 weeks than at 4 weeks, in line with findings in the regenerating avian vestibular epithelia.⁵²

That these developing hair cells may have arisen following stimulation of mitotic activity was suggested by studies performed in vitro using organotypic cultures of explanted utricles from mature guinea pigs and humans⁵⁷; the human material was obtained at operation for acoustic neuromas. The explanted utricles were treated in culture directly with neomycin for 24 hours, then maintained for periods of up to 4 weeks in the continuous presence of tritiated thymidine. Labeled supporting cells scattered within the epithelium were present by 6 days posttreatment and labeling was present in cells putatively identified as hair cells at 4 weeks. Recent morphological studies⁵⁸ have also shown cells with characteristics of developing hair cells in organotypic cultures of guinea pig utricles maintained for 14 days after exposure to gentamicin, suggesting redevelopment of hair cells may occur in vitro. However, in the labeling studies,⁵⁷ the total number of labeled cells in individual cultures was low, much less than the number of hair cells seen by SEM to reappear. In subsequent in vivo studies, BrdU was administered to guinea pigs that were treated with gentamicin to induce hair cell loss.⁵⁹ Labeled supporting cells were found in the utricles 3 days after the end of a 7 day

course of gentamicin treatment. Again, the number of labeled cells was low; between 7 and 38 labeled cells were found in different animals. However, the time of examination was relatively early, at the beginning of the period of progressive hair cell loss when few hair cells have actually disappeared. These findings suggest that hair cell loss in the mammalian vestibular system may induce some proliferative activity and, perhaps, the regeneration of hair cells. There is also some evidence for functional recovery of the peripheral organs of balance after streptomycin treatment over time periods similar to those over which new hair cells appear.⁶⁰

The redevelopment and reappearance of hair cells in vivo in guinea pigs after aminoglycoside-induced losses has subsequently been confirmed.⁶¹ But, after prolonged, continuous administration of tritiated thymidine in vivo, only a few labeled supporting cells could be detected in the sensory epithelia.⁶¹ Because, so far it has been found consistently that the number of labeled cells is considerably less than the number of hair cells seen to be redeveloping, and because label has not yet been detected in cells identified unequivocally as hair cells, there is a possibility that new hair cells in the mammalian vestibular sensory epithelia may arise not through regenerative proliferation, but from the direct transformation of some epithelial cell type into hair cells without going through cell divisions. Such "cell conversion" has been suggested as a mechanism for hair cell production in the embryonic organ of Corti of mice after laser ablations,⁶² as a supplementary mechanism for the production of hair cells in the noise-damaged avian basilar papilla,⁶³ and in the amphibian vestibular sensory epithelia after gentamicin-induced hair cell loss.^{7,64} Cell conversion is now being more widely considered as an alternative means by which new hair cells may be produced, although as yet the evidence for this mechanism is indirect and inconclusive. Alternatively, it has not been conclusively ruled out that some of the redeveloping hair cells appearing in vivo arise through some process in which hair

cells that are damaged, but not killed, repair themselves through a mechanism that assumes some characteristics of normal hair cell development. The evidence currently available would suggest that if this occurs, then damaged hair cells must become dedifferentiated such that they are no longer recognizable as hair cells prior to their redevelopment. There is no precedent for such a process of hair cell damage and self-repair; but this possibility is also under investigation.

In vitro studies have now also provided some evidence for the possibility of stimulating hair cell regeneration in the mammalian organ of Corti.^{64,65} The organs of Corti from 3-day-old rats were maintained in organotypic culture and exposed to neomycin for 48 hours. Morphological assessments using fluorescence microscopy of preparations stained with fluorescently labeled phalloidin, which interacts with actin and thereby labels hair cell stereocilia which are rich in actin, suggested substantial hair cell losses occurred (although other work has shown that hair cell loss in the immature organ of Corti of neonatal mice after neomycin exposure is restricted to the basal turn^{66,67}). Subsequent incubation of the neomycin-injured explants in the presence of retinoic acid, a derivative of vitamin A, and fetal calf serum led to the replacement of nearly all the hair cells after a further 7 days in culture,⁶⁵ but there was considerable irregularity in the arrangement of the hair cells in the recovered organ of Corti. Both the retinoic acid and the serum were necessary to produce recovery; neither alone was effective. The recovery could also be inhibited by an agent, cytosine arabinoside, which blocks mitosis by terminating DNA synthesis prematurely. This was reported as indicative that the replacement hair cells arose following cell division. However, no positive evidence for proliferation, such as hair cell nuclei labeled with tritiated thymidine or BrdU, had been presented by the time this chapter was being prepared.^{68,69} Furthermore, another extensive study failed to find reappearance of hair cells in cultured neonatal rat organ of Corti after neomycin-induced damage,⁶⁹ although it has

been argued that this failure was due to inappropriate culturing conditions.⁷⁰

Subsequent studies⁶⁶ suggested that the factor in serum that is important to the recovery of the injured organ of Corti may be a growth factor, TGF- α (transforming growth factor- α). Growth factors are agents that regulate proliferative activities, and TGF- α is known to stimulate proliferation in some epithelial tissues.⁷¹ Retinoic acid has long been known to be a factor involved in controlling differentiation of cells in many tissues and earlier studies had already established that it may regulate differentiation of hair cells in the immature organ of Corti.⁷² Exposure of the organ of Corti from embryonic mice to retinoic acid at an age just following the normally terminal mitotic events that establish the sensory epithelium, but before differentiation of hair cells and supporting cells is completed, resulted in the appearance of numerous extra hair cells ("supernumerary" hair cells). These arose in the absence of mitotic cell division. This indicates that there are cells in the immature organ of Corti whose differentiation can be influenced by exposure to appropriate factors at the appropriate time.

It is important to stress that the studies of hair cell recovery in the mammalian organ of Corti were performed with explants from neonatal rats. Rats are born deaf, and the organ of Corti of the neonate is immature, becoming fully developed about 3 weeks after birth.⁷³ Neither the inner nor the outer hair cells of the neonate show the specializations of mature mammalian cochlear hair cells. These characteristics do not develop until about 8–10 days after birth,⁷⁴ and prior to this time the hair cells more closely resemble the relatively less differentiated mature hair cells of the vestibular system and the avian basilar papilla. Furthermore, and perhaps significantly, in the immature organ of Corti of the neonate, supporting cells have not begun to undergo the maturational alterations that result in the formation of the large extracellular spaces seen in the mature organ of Corti. In the organ of Corti of the neonate the supporting cells closely appose the bodies of the hair cells^{73,75}

in a manner that resembles the hair cell-supporting cell relationship seen in the mature vestibular system, the avian basilar papilla, and the inner ear sensory epithelia of lower vertebrates.

Triggers for Hair Cell Regeneration and Precursor Cells

The lateral line system of amphibians offers a convenient model for examining the relationship between hair cell loss and regeneration because the neuromasts, the sensory structures of the lateral line that contain hair cells, are located close to the surface and in a number of species can be examined quite easily in living animals. Direct *in vivo* studies of the lateral line of salamanders have shown that hair cell loss stimulates mitosis in supporting cells that surround the hair cells.^{2,76} Individual hair cells were killed by focusing a laser beam at their nuclei and the ensuing events were recorded with time-lapse video. The dead hair cell was extruded from the epithelium and the lesion repaired by expansion of the apical surfaces of adjacent supporting cells. Subsequently mitosis was initiated in one of these supporting cells and successive rounds of division occurred before one of the progeny developed into a new hair cell. In subsequent studies⁷⁷ a laser beam was used to ablate hair cells in cultured explants of the avian basilar papilla. After the "laser surgery" the cultures were maintained in medium containing proliferation marker and labeled supporting cells were identified in the locations of the ablated hair cells.

Studies of the traumatized avian basilar papilla *in vivo* have led to similar conclusions. Following acoustic overstimulation, new hair cells first appear in the basilar papilla 96 hours after the start of exposure regardless of whether traumatizing noise exposure is for 4 or 48 hours³¹ (Figure 1-1). This implies that the regeneration is initiated and continues during the exposure. Likewise new hair cells are apparent in the basal end of the aminoglycoside-damaged papilla while hair cell loss is still progressing^{32,37} (Figure 1-2). Studies of the initiation of proliferative activity confirm this

conclusion. Labeled cells appear in the papilla within the region of the induced lesion and are present within 18 hours of the start of noise exposure.^{78,79} A peak of proliferative activity is seen between 42 and 48 hours from the start of exposure,⁷⁶ corresponding to the period of most rapid hair cell loss. Furthermore, examination of the distribution of BrdU-labeled cells with time from onset of the damaging noise has shown that they appear in a pattern that matches the progression of hair cell loss,⁸⁰ that is after relatively short exposures, a few labeled hair cells were present toward the abneural edge of the papilla; and after increasing exposure times, labeled cells were present in greater numbers at more neural locations. After kanamycin³⁷- or gentamicin⁸¹-induced damage, labeled cells were scattered throughout the papilla along the region of hair cell loss, and could be found as early as 4 days after a single gentamicin injection.⁸¹ These observations confirm earlier suggestions¹² that it is loss of hair cells that stimulates the production of hair cells. Moreover, because after noise exposure labeled cells were found in clusters within the lesion area, it may be that several rounds of division are initiated before repair is completed.⁷⁸ In addition, labeled hair cells appear within the region of the lesion⁷⁸⁻⁸⁰ suggesting that it is the supporting cells that proliferate after trauma-induced hair cell loss. Direct evidence for this has come from light microscope^{46,79} and thin section⁸² observations of mitosis in identified supporting cells of the noise-damaged basilar papilla. The mitosing cells remained attached to adjacent cells at the luminal surface by apparently intact intercellular junctions, but they were rounded; their nuclei, which normally are located toward the base of the cell, had migrated toward the luminal surface; they were detached from the basal lamina underlying the sensory epithelium, which the supporting cells normally contact; and the rounded cell body had lost direct contact with the bodies of the adjacent cells within the epithelium. These features of the mitotic cells seen after noise trauma correspond closely with those observed during the production of hair cells and supporting cells from their common precu-

sors during embryonic development of the avian basilar papilla.⁸³

The observations that during the progressive recovery from trauma in the avian vestibular system, proliferation labels appear in supporting cell nuclei before they appear in hair cell nuclei, complement these findings from the lateral line and the avian basilar papilla that directly demonstrate proliferation among the supporting cell population. Altogether, they suggest a common process of repair in which hair cell loss stimulates proliferation of undamaged nonsensory cells in the epithelium and subsequent differentiation of immediate or succeeding progeny into hair cells. The non-sensory cells within these epithelia are usually classified as a single population of supporting cells. Thus, potentially all of these cells may possess the capability to be stimulated to enter the cycle. However, some evidence has been presented to suggest that in the inner ears of fish there may be a resident sub-population of nonsensory cells, separate from the supporting cells, that are proliferation competent and are the precursors for the stimulated production of new hair cells.⁸⁴ There is no evidence indicating subpopulations of nonsensory cells in the avian or mammalian inner ear sensory epithelia, but there is very little published work in which this possibility has been examined.

Processes of Hair Cell Loss and Epithelial Repair Following Injury

The loss of a hair cell will potentially create a lesion through the luminal surface of the epithelium. In inner ear epithelia, the apical surface is bathed in a potassium-rich fluid, endolymph, whilst the cell bodies are bathed in sodium-rich fluid. The loss of a hair cell might be expected to disrupt the permeability barriers that are normally present to separate these two fluids. By allowing the entry of endolymph into the *corpus* of the epithelium, the cell bodies would be exposed to an abnormal and potentially damaging environment that could disturb tissue homeostasis. This is likely to impede any repair or regeneration events.

After laser ablation of hair cells in lateral line neuromasts,² noise trauma to the basilar papilla,¹⁰ and aminoglycoside-induced injury in the avian basilar papilla and the vestibular system of fish, amphibia, and birds,^{6,7,52} hair cells are expelled from the apical surface of the epithelium. Sections of the noise-damaged basilar papilla have shown that the extruded cells contain nuclei, indicating that injured cells are expelled intact.¹⁰ Lost hair cells are initially replaced by expansion of the adjacent supporting cells to effect a repair.

Examination of the avian basilar papilla after noise trauma has suggested that the loss of hair cells and their replacement by expansion of adjacent supporting cells is achieved without causing any obvious disruption of the continuity of the apical surface nor of the integrity of tight junctions,⁷⁹ which maintain the permeability barriers at that surface. Furthermore, in addition to expulsion of cells from the apical surface, degeneration of hair cells within the noise-damaged basilar papilla has been observed.⁴⁶ The nuclei of these degenerating cells showed fragmentation and condensation of their chromatin, morphological features of apoptosis, a programmed cell death phenomenon. Apoptosis occurs "naturally" in many normal and developing tissues to regulate cell numbers, but also after some cytotoxic challenges as a means of removing cells without disrupting tissue integrity.⁸⁵⁻⁸⁷ In the mammalian organ of Corti, after aminoglycoside injury, it has also been shown that hair cell bodies degenerate within the epithelium showing some morphological features similar to those of apoptosis.⁸⁸ The hair cells appear to rupture. An apical fragment is retained in the reticular lamina, the junctions with adjacent supporting cells apparently remaining intact⁸⁹ until supporting cells expand into the region of the degenerated cell body and close the lesion by the formation of new sealing junctions.⁹⁰ In the mammalian vestibular sensory epithelia a similar process of repair seems to occur; hair cells degenerating within the epithelium show some morphological features of apoptosis.⁵⁰ But loss of hair cells from this tissue additionally occurs through an apparently active extrusion of intact cells from

the surface of the epithelium. In this latter case, the expelled hair cell remains attached at the surface until supporting cells have sealed the lesion site. Although precise details of the manner of hair cell loss have not been examined in lower vertebrates, these observations indicate that hair cell loss after injury may be a controlled process in which supporting cells play a key role in effecting repair with a minimum of disruption of tissue integrity. It has been argued that in other epithelial tissues where regeneration of cells is known to occur after cytotoxic injury, for example pancreatic acinar and intestinal crypts, maintenance of tissue integrity is crucial to allow subsequent regeneration processes to operate.⁹¹⁻⁹³ It seems likely, therefore, that a necessary first step in the process leading to hair cell regeneration in the damaged sensory epithelia is an effective, controlled repair of the lesions by undamaged supporting cells.

Following hair cell loss in the neuromasts of the lateral line, it has been reported that macrophages infiltrate the epithelium, presumably as part of an inflammatory response stimulated by the damaged cell.⁷⁶ Macrophages are thought to release mitogens, agents that stimulate cell proliferation, and in certain cell types, such as muscle, regeneration appears to be dependent upon the presence of macrophages.⁹⁴ Thus, it has been suggested that macrophages may have a role in stimulating hair cell regeneration. In further support of this, recent *in vitro* studies of the avian basilar papilla maintained in organotypic culture have shown leukocytes to be attracted to the apical surface of the epithelium specifically at the site of an induced lesion.⁹⁵ However, in studies of the gentamicin-damaged mammalian vestibular system, whilst macrophages have been observed within the epithelium very occasionally, this was most uncommon.⁵⁰ Furthermore, there is no reported evidence for macrophage infiltration into the injured sensory epithelia of the inner ears in birds, amphibia, or fish *in vivo*. Indeed, it is noteworthy that an inflammatory response does not appear to accompany hair cell loss in these epithelia. This could be related to the modes

by which damaged hair cells are lost. The expulsion of hair cells intact from the apical surface of the epithelia may reduce the likelihood of the release of cytoplasmic material, which could provoke inflammatory responses, from the damaged cell into the epithelium. Apoptosis, during which the plasma membranes of dying cells remain intact, is regarded as a means of deleting cells without stimulating exudative inflammation and this has been cited as a factor in allowing regenerative processes to be rapidly initiated following epithelial injury.^{85,92} However, although there is little conclusive evidence for a role for non-epithelial cells in stimulating proliferative activity leading to hair cell regeneration, this has not been examined in any detail. At present, it appears that proliferation is triggered by factors within the resident population of epithelial cells.

Regardless of the mechanisms by which repair is effected, following damage to the sensory epithelium, some supporting cells lose contact with hair cells and then reestablish new contacts and junctional connections between themselves. It has been suggested that these changes in supporting cell contacts provide the triggers for stimulating activity in supporting cells resulting in hair cell regeneration.¹⁰ The intercellular junctions between two supporting cells at the luminal surface of the epithelium show distinct morphological differences from those between a hair cell and a supporting cell.^{87,96-98} In addition, adjacent supporting cells form gap-junctional connections (regions of direct cell-cell communication) between each other,⁹⁶⁻⁹⁸ whereas, at least in birds and mammals, the hair cells do not form gap junctions with their adjacent supporting cells.^{96,99,100} When supporting cells expand to effect repair of lesions, the junctions formed at the luminal surface are the same as those normally seen between two adjacent supporting cells, and formation of new gap junctions between the newly adjacent supporting cell bodies has been observed during repair in the organ of Corti after aminoglycoside-induced hair cell loss,⁸⁹ and, in very recent studies, in the avian basilar

papilla following acoustic overstimulation (A. Forge and Y. Raphael, unpublished data, 1995). The supporting cells, therefore, normally have "heterologous" junctional connections, some with other supporting cells, and others with hair cells. Hair cells on the other hand have "homologous" junctions with all their surrounding supporting cells. Upon loss of a hair cell, the intercellular junctions of supporting cells become uniform.

It has been argued that the normal architecture of the inner ear sensory epithelia in which a hair cell is separated from its neighbors by intervening supporting cells to form the characteristic mosaic of cells seen at the apical surface, results during development from "lateral inhibition."¹⁰¹ This hypothesis suggests that after the mitotic production of the precursors which will go on to differentiate into hair cells and supporting cells that start to differentiate as hair cells would exert an inhibitory influence on those cells with which they are in direct contact preventing these neighbors from following the same developmental fate as themselves. The inhibited cells would then develop into supporting cells. Other cells would develop into hair cells if they do not contact a cell that is also differentiating into a hair cell, and so on until the epithelium is formed. It is conjectured that this inhibitory effect may be exerted through the junctional contacts where various intercellular signaling molecules are known to reside. Such lateral inhibition through proteins associated with junctional complexes has been implicated in the control of differentiation in peripheral sense organs in invertebrates as well as vertebrates.⁹⁹ Moreover, genes that are involved in the control of sense organ development in the fruit fly *Drosophila*, have homologues that are expressed in the developing inner ear of chicks,¹⁰² suggesting biologically conserved molecular mechanisms are associated with sense organ development. Recent studies of the developing vestibular system of chicks, however, have indicated that in the avian utricle, supporting cells differentiate before hair cells, whereas in the saccule, the hair cells differentiate first.¹⁰³ This suggests that the generation of

the epithelial mosaic may also require a positive control in which, when a certain number of supporting cells surround an as yet uncommitted cell not in contact with a differentiating hair cell, that uncommitted cell is stimulated to differentiate as a hair cell. Nevertheless, the development of the sensory epithelium may depend upon a relatively simple mechanism in which each cell provides signals that control differentiation to its direct neighbors.

It is a mechanism similar to this that has been proposed as triggering the production of new hair cells after the loss of hair cells and, thus, the "heterologous" junctions of the supporting cells.^{10,76} The removal of hair cell derived inhibitions (and/or the triggering of supporting cell-mediated stimulation) would activate the supporting cells to produce replacement hair cells. As these begin to differentiate, they would influence the development of adjacent cells in the same way as occurs during normal development so that a "normal" epithelium is reestablished.⁷⁴ The predictions from this hypothesis are that the signals necessary for regenerating hair cells after damage are present locally within the sensory epithelium itself and that the molecules involved are probably bound at the cells' surfaces since activation is suggested to be initiated by direct contact between cells. These ideas are consistent with the observations that hair cell loss stimulates the regenerative response, and the usual observation that new hair cells are produced only at sites of lesioning. This would imply there is little or no "action at a distance" mediated by a diffusible molecule released into the epithelium. However, there are reports of new hair cells occasionally appearing in the basilar papilla after acoustic trauma at sites remote from the location of the defined lesion.⁴⁶ It is difficult to be certain that there was indeed no hair cell loss at the site at which the isolated new hair cells were observed, but if this were the case, then this observation would suggest that supporting cells can produce new hair cells without losing contact with hair cells, and that there may be factors involved in stimulating hair cell production that are diffusible.

In Vitro Studies of Hair Cell Regeneration

The use of explants of inner ear tissues maintained in organotypic culture has already been described in connection with the demonstration of proliferation in the adult mammalian vestibular system after aminoglycoside treatment, and the apparent stimulation of hair cell production after drug-induced damage in the immature mammalian organ of Corti. Such *in vitro* systems offer a convenient means for examination of tissues under controlled, easily manipulated conditions, and they are being used increasingly to attempt to identify factors involved in hair cell regeneration.

The avian vestibular sensory epithelia have been the preferred tissues for many *in vitro* studies because these epithelia appear to be continually producing hair cells and the production can be increased ("up-regulated") after aminoglycoside-induced damage. It has been demonstrated that the normal proliferative activity, assessed by culturing tissues in the presence of tritiated thymidine to label the nuclei of proliferating cells, occurs in a medium devoid of serum.^{104,105} Further, by explanting utricles that had been damaged *in vivo* with streptomycin, it has been demonstrated that an increase in proliferation and differentiation of progeny into hair cells also occur *in vitro*.¹⁰⁵ Because serum contains many growth factors and accessory agents of importance to cell production, the continuation of proliferation in its absence argues that all the factors necessary for proliferation, and perhaps differentiation too, are present within the tissue itself, and that no external source of growth factors is necessary. This would be consistent with the hypothesis that loss of hair cells and reorganization of the epithelium provide sufficient triggers for hair cell regeneration. Other studies in which several different explants were maintained together in co-cultures¹⁰⁶ have suggested that some of these proliferation-controlling factors produced by the sensory epithelium itself may be soluble and able to "act at a distance." Cultures of *in vivo* streptomycin-damaged utricles, in which proliferation is up-regulated fol-

lowing the injury, with undamaged utricles, in which there is a lower basal rate of proliferation associated with cell turnover, led to increased numbers of proliferating (tritiated-thymidine labeled) cells in the undamaged tissue compared with controls. Interestingly, culturing of several undamaged utricles together led to an apparent decrease in the number of proliferating cells compared with undamaged tissue cultured in isolation. This suggests there may be soluble inhibitory factors produced by normal sensory epithelia that depress cell proliferation, although other explanations are possible, for example increased death of proliferation-competent cells.

In other *in vitro* studies of undamaged tissue, the effects of added growth factors have been examined in efforts to identify possible agents that the epithelium itself might be using to regulate the process of cell proliferation. In avian utricles, it was found that addition of insulin-like growth factor-1 (IGF-1) to culture medium led to increased numbers of labeled nuclei in comparison with a number of other growth factors tested, including epidermal growth factor (EGF) and TGF- α that had no effect.¹⁰⁵ In mammalian utricles, TGF- α alone and in combination with insulin, and EGF in combination with insulin appeared to induce proliferative activity.¹⁰⁶ Furthermore, the addition of TGF- α to culture medium after neomycin damage to mammalian utricles *in vitro*, has been shown to stimulate the production of significantly greater numbers of cells labeled with proliferation markers than are produced in its absence,¹⁰⁷ although the total number of labeled cells was still relatively small. As mentioned above, TGF- α has also been suggested to be a factor involved in the production of hair cells in the neonatal mammalian organ of Corti after aminoglycoside injury.⁶⁴

These *in vitro* studies represent only a beginning. It needs to be stressed that studies of growth factors so far have shown only the stimulation of proliferation and not the production of hair cells. As discussed in detail by Cotanche et al.,²⁰ several different factors alone or in combination are likely to be involved *in vivo* in triggering normally mitot-

ically quiescent cells of the inner ear sensory epithelia to reenter the cell cycle; in guiding them through the cycle, and in controlling subsequent differentiation. It is likely, therefore, that further investigation of those factors involved in stimulating hair cell regeneration will produce a minefield of misinterpretation, false dawns, and controversy, but on the other side may well lie the clues to wisdom.

Conclusion

The phenomenological similarities in processes of repair and recovery in the sensory epithelia of lateral line organs and those of the inner ear in fish, amphibia, and birds suggest common mechanisms conserved across these vertebrate classes. In addition, current evidence indicates that the potential to replace hair cells lost after injury may also exist in mammals. The increasing ability to apply contemporary techniques of cell and molecular biology to inner ear sensory tissues provides means to gain an understanding of the bases of these phenomena and how they are controlled. It seems reasonable, therefore, at least to consider the eventual possibility of attempting to stimulate effective repair of the traumatized mammalian organ of Corti.

During trauma-induced hair cell loss in the organ of Corti, supporting cells immediately close the lesions caused in a manner that is similar to that seen in their counterparts in other inner ear sensory epithelia. This indicates that these cells have retained the ability to sense and to respond effectively to hair cell injury, the apparent first, necessary step in the processes that leads elsewhere to hair cell regeneration. However, following this initial repair in the organ of Corti, instead of producing new sensory cells, the epithelium undergoes a prolonged and progressive "dedifferentiation" in which neural elements degenerate and which ultimately results in the replacement of the specialized sensory epithelium with nonspecialized, squamous-like epithelial cells.^{108,109} Presumably there are good biological reasons why the mammalian organ of Corti has evolved to respond to in-

jury through a controlled degeneration rather than by restoration of the sensory epithelium. Stimulation of regeneration would represent an unnatural response of this tissue, the consequences of which are difficult to predict. Certainly it seems unlikely that the highly organized, specialized organ of Corti could be re-created in a previously "deaf" ear. Furthermore, even if hair cells were replaced they would need to become reinnervated appropriately. The regenerative capacities of cochlear innervation are not known. How to restore innervation to the damaged organ of Corti in a useful pattern represents another set of challenges that need to be addressed.^{110,111}

On the other hand, experience with cochlear implants indicates that the sensory input that the higher neural centers require in order to provide useful auditory information need not be very sophisticated. Thus, it may only be necessary to replace a small number of hair cells and rely upon the existing remnant neural elements to enable some restoration of hearing to a previously deaf ear. To discover the feasibility of that goal makes continued investigation of the mechanisms of hair cell regeneration worthwhile.

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Chapter 2

The Effects of Acoustic Trauma, Other Cochlear Injury, and Death on Basilar-Membrane Responses to Sound

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In pioneering experiments, Nelson Kiang and associates showed that hair cell loss induced by kanamycin raises the thresholds and reduces the frequency selectivity of auditory-nerve fiber responses to tones.¹ These initial findings were subsequently extended in investigations of cochlear dysfunction induced by acoustic trauma,² chronic or acute administration of ototoxic chemical agents,^{3,4} or anoxia.^{5,6} For as long as the vibrations of the basilar membrane could plausibly be taken as linear, poorly frequency tuned, and largely invulnerable to physiologic insults, the effects of such insults on auditory-nerve function were usually interpreted as reflecting injury to a hypothetical second filter located in the organ of Corti, that sharpens the frequency tuning of the first filter, the basilar membrane.⁷

The second filter concept held sway until the 1980s. Surprisingly this concept retained currency for longer than a decade after William Rhode demonstrated that the squirrel monkey basilar membrane responds to sounds with vibrations that grow nonlinearly at the characteristic frequency (CF) and are physiologically labile.⁸⁻¹⁰ Unfortunately, for a number of years Rhode's discovery was viewed with a great degree of skepticism, largely as a result of the failure of several investigations (including one by Rhode himself) to find basilar-membrane nonlinearities in species other than the squirrel monkey.^{7,11}

Eventually, sensitive and sharply tuned

basilar-membrane responses that grow nonlinearly at the CF and are physiologically vulnerable were reported in three mammalian species, namely guinea pig,¹²⁻¹⁴ chinchilla,¹⁵ and cat.¹⁴ Although the presence of mechanical basilar-membrane analogs of auditory-nerve response properties by itself does not rule out the existence of the second filter, it does render it theoretically unnecessary. Significantly, several investigations have shown that cochlear insults that damage auditory-nerve function in a frequency-specific manner also alter basilar-membrane vibrations. The present work reviews such investigations, surveying the effects of acoustic overstimulation, furosemide, quinine, death, and experimentally induced cochlear trauma upon basilar membrane responses to single tones, clicks, and pairs of tones. The mechanical effects are consistent with the idea that the alterations of basilar-membrane vibration causally determine the neural effects. However, the mechanical effects of cochlear insults also imply that the organ of Corti influences the response to sound of the basilar membrane.

Effects of Death and Experimentally Induced Cochlear Trauma: Seminal Findings

Noting that basilar-membrane response nonlinearities are abolished by death, Rhode suggested that, far from being quirky results of

the Mössbauer methodology, they are actually central to normal cochlear function.⁸⁻¹¹ At the time, however, Rhode was unable to prove this contention because his basilar-membrane recordings were not accompanied by independent controls (e.g., auditory-nerve thresholds).

Perhaps most confusingly, when independent neural controls were obtained, the results appeared to contradict both Rhode's experimental findings and conclusions. Evans and Wilson⁷ recorded from individual auditory-nerve fibers in cat cochleae from which they also recorded basilar-membrane responses. The basilar-membrane vibrations were linear and broadly frequency tuned. In contrast, responses from fibers seemingly innervating the cochlear region from which mechanical vibrations were obtained were sensitive and sharply tuned. At face value, these findings not only refuted the idea that the nonlinearity is a normal property of basilar-membrane vibration, but also implied the existence of a second filter. In light of later results, today one supposes that, in fact, the basilar-membrane recordings were from damaged cochlear regions (perhaps injured by removal of perilymph and/or by electrical currents from the capacitive probe) and the neural recordings were obtained from fibers innervating nearby relatively normal regions.

Effects of Death and Experimentally Induced Cochlear Trauma: An Update

The causal linkage between basilar-membrane nonlinearity and the sensitivity and sharpness of tuning of cochlear responses became generally accepted only when Sellick et al.¹² and Robles et al.¹⁵ showed that in guinea pigs and chinchillas mechanical sensitivity, frequency selectivity, and nonlinearity are highly correlated with the thresholds of compound action potentials (CAPs). The correlation between neural and mechanical sensitivity is well illustrated in Figures 2-1 and 2-2, taken from the classic work of Sellick et al.¹² Initially, sharply tuned and sensitive basilar-membrane responses were recorded when tone pips, with frequency equal to the mechanical CF, evoked

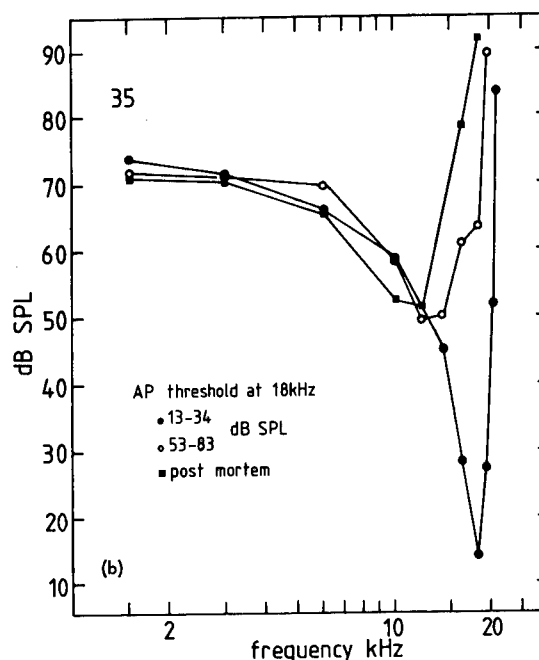


Figure 2-1 Correlation between auditory-nerve and basilar-membrane sensitivity and frequency tuning. Basilar-membrane isoresponse functions were initially recorded in a guinea pig while the cochlea was healthy (filled circles), as indicated by low thresholds (13–34 dB SPL) for the compound action potential elicited by tone pips with frequency equal to the CF of the mechanical recording site. When the cochlea deteriorated, as indicated by elevated neural thresholds, the sensitivity of basilar-membrane responses was reduced. Reproduced, with permission, from figure 15B of Sellick et al.¹²

low-threshold CAPs; as time passed, cochlear function deteriorated, resulting in loss of mechanical sensitivity and frequency selectivity and, simultaneously, elevation of neural thresholds.

The sensitivity of basilar-membrane responses is closely connected to an intensity-dependent nonlinearity. Figure 2-3 displays the gains (velocity magnitude divided by pressure) of basilar-membrane responses to tones and clicks, as a function of frequency. The solid lines, representing vibrations recorded by means of laser velocimetry in a relatively healthy chinchilla cochlea, show that response sensitivity (i.e., gain) at near CF frequencies varied strongly with stimulus inten-

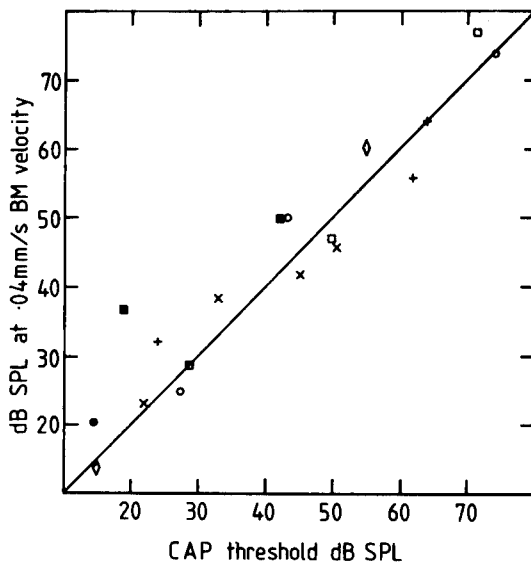


Figure 2-2 Correlation between auditory-nerve threshold and basilar-membrane sensitivity. The abscissa indicates the thresholds of CAPs evoked by tone pips with frequency equal to the CFs of many basilar-membrane recordings sites. The ordinate indicates the sound pressure level (dB re 20 μ Pa) required for eliciting a basilar-membrane velocity of 0.04 mm/second at CF. Reproduced, with permission, from figure 9 of Sellick et al.¹²

sity. The change in sensitivity, with the largest gains corresponding to the lowest stimulus levels, corresponds to a compressive input-output law (1 dB increments in stimulus intensity eliciting response growth smaller than 1 dB). At frequencies distant from CF, on the other hand, response sensitivity did not vary with stimulus intensity; rather, responses grew linearly. Figure 2-3 also displays gain functions (dashed lines) for the insensitive and broadly tuned responses measured postmortem. The nearly complete overlap among responses to click stimuli at 87, 97, and 107 dB sound pressure level (SPL; upper panel) demonstrates that, in clear contrast with the results in the live cochlea, postmortem responses were linear at all frequencies.

Figure 2-4 shows the phases of some of the responses to clicks depicted in Figure 2-3, relative to stapes inward motion, as a function of stimulus frequency.¹⁶ The change of phase with frequency has three components. At

the lowest frequencies, the phase versus frequency curves are essentially flat. At intermediate frequencies (3–8 kHz), the phase curve has a shallow slope, corresponding to a group delay of about 125 microseconds, which is the time that it takes a disturbance to propagate from the oval window to the basilar-membrane recording site, nearly 3.5 mm away. In the region around CF, the phase versus frequency slope is much steeper, with an average group delay of 520 microseconds for high level responses. Corresponding to the nonlinear growth of responses at near CF frequencies, phases varied at these frequencies as a function of stimulus intensity: for frequencies below CF, responses to high-

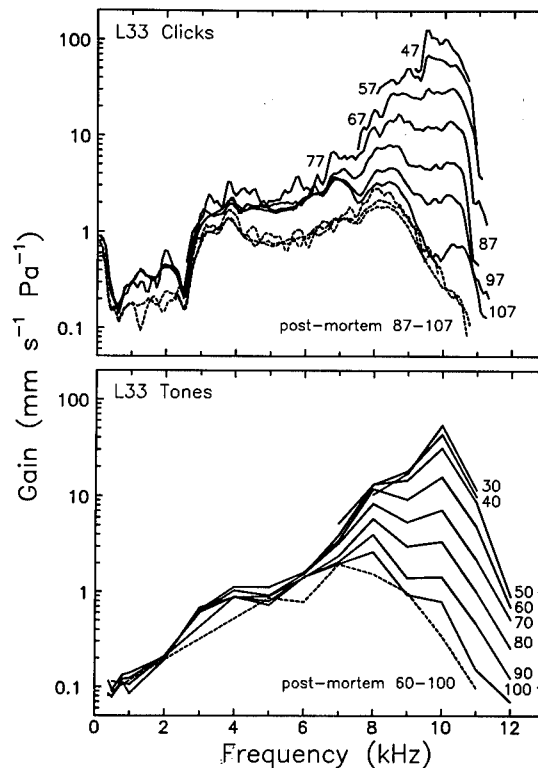


Figure 2-3 Velocity-gain spectra for responses to clicks (top panel) and tones (bottom panel) in the same cochlea. The gain spectra were computed by dividing velocity spectra by the stimulus pressure (indicated, in dB SPL, as the parameter). Continuous lines signify responses in the relatively normal cochlea and dashed lines indicate postmortem responses. Reproduced, with permission, from figure 3 of Ruggero et al.¹⁶

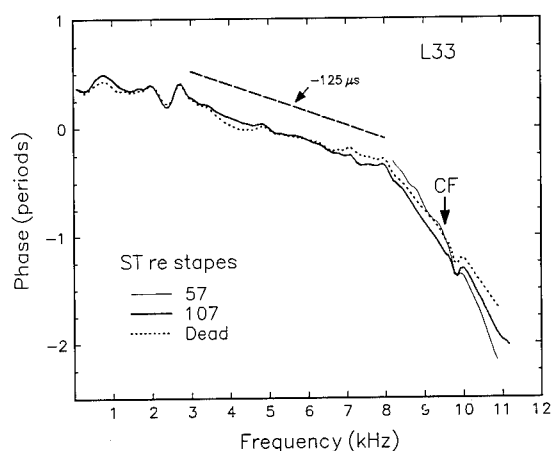


Figure 2-4 Phases of basilar membrane responses to clicks in a live cochlea (continuous lines) and postmortem (dotted line). Phases, computed by Fourier transformation, correspond to basilar membrane displacement toward scala tympani relative to inward stapes displacement. The responses of clicks were digitally deconvolved to reduce the effect of irregularities in the stimulus frequency spectrum. The thick continuous line represents responses to clicks with a peak SPL of 107 dB; the thin line represents responses to 57 dB clicks. The dashed line has a slope of -125 microseconds. The down-pointing arrow indicates the characteristic frequency. Reproduced, with permission, from figure 2 of Ruggero.³⁸

intensity stimuli lagged those to lower level stimuli; for stimulus frequencies above CF, responses to intense stimuli tended to lead low-level responses. The (nonlinear) phase changes, however, were small in comparison with the overall phase lag at frequencies near CF. Similarly, and in keeping with the above noted resemblance between normal responses measured at high stimulus intensities and postmortem responses, the phase changes accompanying death were relatively minor. The near CF group delay of postmortem phases was only slightly smaller (by 50 microseconds) than that of the normal responses to high intensity stimuli.

The postmortem phases of Figure 2-4 agree with Rhode's¹⁰ initial description of the effects of death upon basilar membrane response phases in that both sets of data show a diminished postmortem near CF group delay.

Rhode also reported that death induces relative phase lags at CF and substantial *increases* in group delay (phase slope) at frequencies below CF. Although not apparent in Figure 2-4, other recordings in our laboratory agree with Rhode's findings in that death (or other cochlear trauma; see Figures 2-6 and 2-11, below) tends to cause increased phase lag at CF. However, we have never seen significantly increased phase lags or group delays at frequencies far below CF, at which the basilar membrane vibrates linearly. We suggest that the large postmortem phase lags reported by Rhode,¹⁰ even at low frequencies, represent drastic changes in the passive properties of the cochlear partition, perhaps including an increase in its elasticity.

Effects of Acoustic Overstimulation

Figure 2-5 illustrates how basilar-membrane response magnitudes are affected by exposing the ear to intense tones with a frequency about $\frac{1}{2}$ octave below CF. The left panel of Figure 2-5 makes evident the marked nonlinearity of preexposure responses to clicks at spectral frequencies near CF: the gain peaks are largest for the weakest stimuli and become progressively smaller with increasing click intensity. Following a 4-minute exposure to a 7 kHz, 100 dB SPL tone, responses to clicks were reduced in magnitude and linearized (right panel). Acoustic overstimulation reduced the gain of responses to the weakest clicks by about 20 dB, leaving responses to the most intense clicks essentially unaffected. The gain reduction was frequency specific in that CF responses were affected the most and responses at frequencies well below CF were essentially unchanged. As a result, the apparent CF was somewhat lowered by the acoustic overstimulation. The nearly complete overlap between the gain functions at all frequencies indicates a proportional (i.e., linear) growth of responses with stimulus level.

The effects on response magnitude of exposure to intense tones illustrated in Figure 2-5 are generally representative of our findings in several chinchilla cochleae¹⁷ and of isolated results for cats and guinea pigs.^{14,18} The mag-

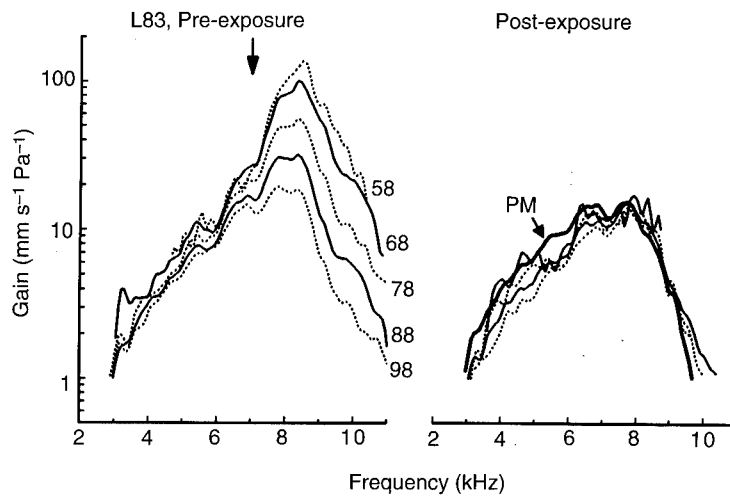


Figure 2-5 The effect of acoustic overstimulation on basilar-membrane responses to clicks. The left and right panels, respectively, show gain spectra (velocity spectra normalized to sound pressure level) for responses to clicks preceding and following a 4 minute exposure to a 7 kHz, 100 dB SPL tone. The parameter in the left panel indicates peak click pressure. Postexposure responses were elicited by clicks with peak pressures of 68–98 dB. The thick line in the right panel is the postmortem (PM) gain function. Reproduced, with permission, from figure 3 of Ruggero et al.¹⁷

nitide effects were invariably accompanied by phase lags confined to frequencies near the CF (Figure 2-6). In this respect, our results and those of Cooper and Rhode¹⁴ in the cat, contrast with phase *leads* reported by Patuzzi et

al.¹⁸ after acoustic overstimulation in one guinea pig cochlea.

What processes are responsible for the basilar-membrane effects of acoustic trauma? At present the most reasonable hypothesis is

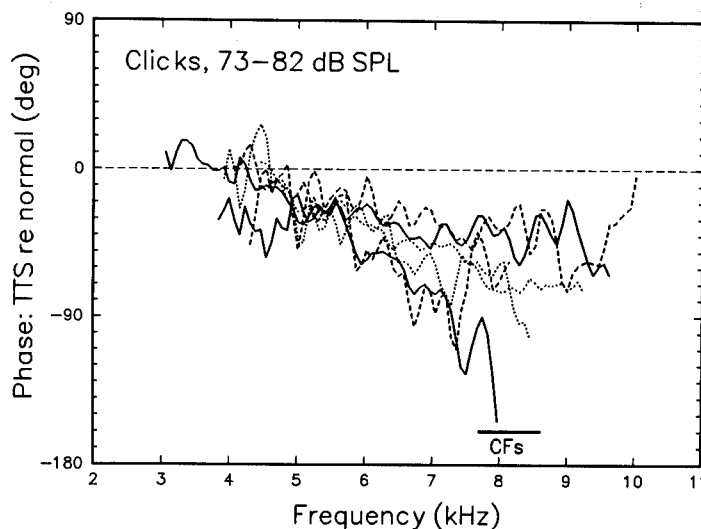


Figure 2-6 The effect of acoustic overstimulation on basilar-membrane response phases. The net changes in phase spectra are shown for six cochleae. Each cochlea is represented by a single curve computed from responses to clicks with peak levels of 73–82 dB. Negative phases indicate phase lags. Reproduced, with permission, from figure 4 of Ruggero et al.¹⁷

that acoustic trauma primarily disrupts mechanical to electrical transduction in outer hair cells, secondarily reducing the mechanical assist that outer hair cells presumably provide to basilar-membrane vibrations. The best evidence for this idea is that acoustic trauma reduces the magnitude of outer hair cell receptor potentials elicited by low-frequency tones in high CF cochlear regions,¹⁹ where basilar-membrane vibrations are linear and unaffected by acoustic trauma at stimulus frequencies well below CF. Since basilar-membrane vibrations are linear and unaffected by acoustic overexposure at stimulus frequencies well below CF, the alteration of outer hair cell receptor potentials implies that acoustic trauma acts at a stage of signal transformation located central to basilar-membrane vibration.

The effect of acoustic overstimulation has also been studied in the apical region of an *in vitro* preparation of the guinea pig temporal bone.²⁰ Even preceding acoustic overstimulation, the *in vitro* preparation lacks the sensitivity and CF-specific nonlinearities that characterize normal basilar-membrane vibrations at the cochlear base. Surprisingly, overstimulation apparently caused *increases* in the vibratory response of the organ of Corti. This unique finding is difficult to reconcile with either basilar-membrane measurements in relatively healthy cochleae or with the extensive literature on anatomical, neurophysiological, and psychophysical correlates of acoustic trauma.^{21–23}

Vulnerability of Two-Tone Suppression and Distortion in Basilar-Membrane Responses

Nonlinearities in auditory-nerve responses to two tones, namely two-tone rate suppression and the generation of the cubic difference tone ($2f_1 - f_2$), are labile to cochlear injury.²⁴ Because both two-tone suppression and distortion probably originate in corresponding basilar-membrane phenomena,^{13,25–28} it is reasonable to expect that the vulnerability of the auditory-nerve nonlinearities is also rooted in basilar-membrane behavior. That this is indeed the case is illustrated in Figures 2-7 and 2-8. Figure 2-7a shows velocity–intensity functions

obtained in one chinchilla basilar membrane for a near CF probe tone alone and in the presence of a low frequency suppressor tone, before and after death of the animal. While the chinchilla was alive, suppression (measured as a horizontal shift along the intensity axis) reduced responses to low- and moderate level CF tones by about 15 dB; for intense CF tones, however, suppression was essentially nil. The intensity dependence of suppression amounted to a linearization of the input–output characteristic. Death did not alter the response to the suppressor tone (not shown) but produced a drastic loss of sensitivity in the CF tone intensity function at low and moderate stimulus levels, essentially abolishing both the intensity-dependent compressive nonlinearity and two-tone suppression. A clear relationship can be demonstrated between the physiological state of the cochlea and the magnitude of measurable mechanical suppression. In one study, cochlear deterioration due to surgical trauma was estimated by determining the SPL required to elicit a 100 $\mu\text{m}/\text{second}$ basilar-membrane response to a near CF probe tone. When this measure of cochlear injury was plotted against suppression magnitude (Figure 2-7b), a strong correlation between the variables was evident: suppression magnitude was large in relatively normal cochleae and small in damaged cochleae.

Figure 2-8 shows that acoustic trauma reduces basilar-membrane responses to $2f_1 - f_2$ distortion products in a frequency-specific manner.²⁹ When a chinchilla cochlea was stimulated with a pair of tones with frequencies f_1 and f_2 , basilar-membrane vibrations contained additional components, among them a prominent one with frequency equal to $2f_1 - f_2$ (8 kHz = CF, in this case). Upon overstimulation for 4 minutes with a 9.5 kHz, 100 dB tone, responses to the $2f_1 - f_2$ distortion product were reduced by at least 13 dB at moderate primary-tone levels; responses to CF tones were diminished by 4 dB or less. The data of Figure 2-8 constitute a mechanical counterpart of, and an explanation for, psychoacoustical observations in humans indicating that the $2f_1 - f_2$ distortion product is abolished or reduced by threshold

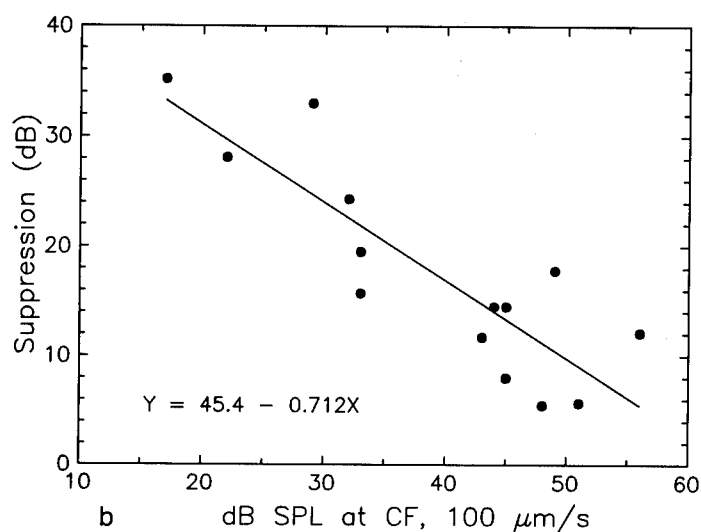
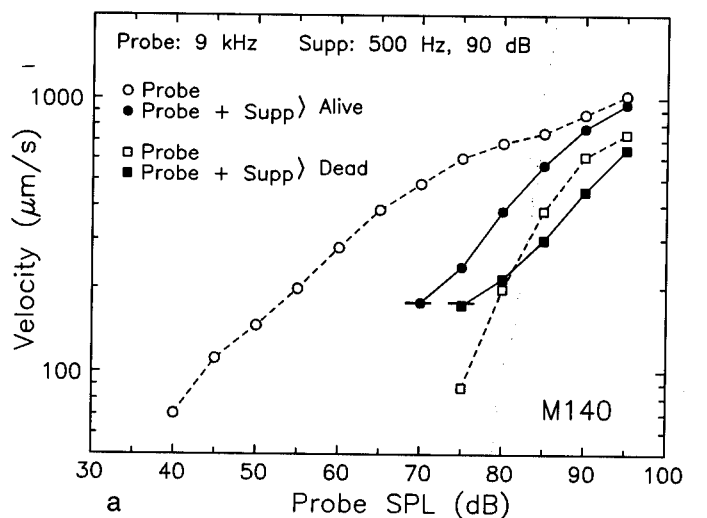


Figure 2-7 The effect of death or surgical trauma on mechanical two-tone suppression. **(a)** Effects of death upon basilar-membrane suppression. Velocity-intensity functions were obtained (\circ) before and (\square) after death in one chinchilla. In each condition, intensity functions are shown for the probe tone (9 kHz) presented alone (open symbols, dashed lines) and for the probe accompanied by a 500 Hz, 90 dB SPL suppressor tone (solid symbols, solid lines). **(b)** Dependence of the magnitude of suppression on basilar-membrane response sensitivity. Data are shown for 14 chinchilla cochleae in which above CF suppressors were used. The abscissa indicates the SPL required for a CF tone to elicit a 100 $\mu\text{m/s}$ response. The ordinate indicates the magnitude of suppression caused by 80 dB above CF suppressors. Panels **(a)** and **(b)**, respectively, reproduced with permission from figures 12 and 11 of Ruggero et al.²⁷

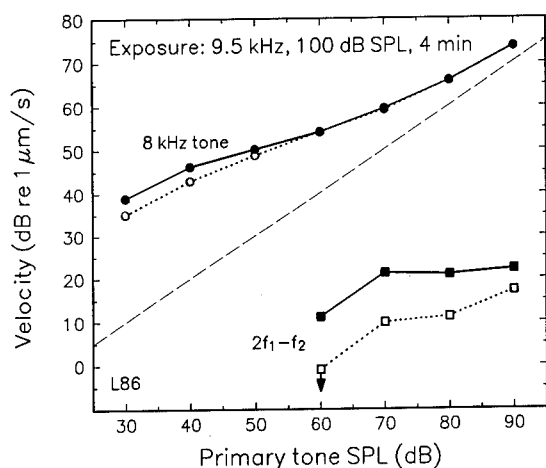


Figure 2-8 Frequency selective reduction of basilar-membrane distortion products by acoustic overstimulation. When an initially sensitive cochlea was stimulated with equal level, two-tone stimuli with frequencies such that $2f_1 - f_2 = \text{CF}$ (8 kHz), the basilar-membrane response included the corresponding distortion product (■). After subjecting the cochlea to overstimulation with an intense tone (10.6 kHz, 100 dB SPL, 4.5 minutes), the $2f_1 - f_2$ distortion products were substantially attenuated (□). In contrast, the responses to CF tones were reduced only slightly (○). (From Robles et al.²⁹)

elevations at frequencies equal to either $2f_1 - f_2$ or, most significantly, f_1 or f_2 .³⁰ Such psychoacoustical data have been interpreted as evidence that the $2f_1 - f_2$ distortion product arises near the cochlear site with CF equal to f_1 or f_2 and then travels to the cochlear site with CF equal to $2f_1 - f_2$, where it is analyzed and detected. Figure 2-8 provides almost unassailable support for this interpretation.

Effects of Quinine

Quinine is an ototoxic drug that reversibly raises auditory thresholds and can induce tinnitus. Quinine elevates auditory-nerve thresholds,³¹ reduces the magnitude of cochlear microphonics,^{31,32} alters otoacoustic emissions,^{32,33} and induces changes in the morphology of the subsurface cisternae of outer hair cells³⁴ but quinine does *not* reduce the endocochlear potential.³¹ Although its mode of action is uncertain, the cochlear ef-

fects of quinine are consistent with a direct effect on the outer hair cells that may secondarily result in alterations of basilar-membrane mechanics. We have recently obtained evidence that quinine, in fact, reduces the magnitude of basilar-membrane vibration.³⁵ Figure 2-9 allows side by side comparison of the effects of an intravenous injection of quinine upon basilar-membrane intensity-velocity functions for responses to a CF- and a low-frequency tone. Quinine caused a 15 dB reduction in the sensitivity of responses to low level CF tones, but did not appreciably change responses to intense CF tones. Responses to low frequency tones were unaffected at all levels measured. As seen above for the effects of death or acoustic overstimulation (Figures 2-3, 2-5, 2-7a), the net effect of quinine was to linearize basilar-membrane vibrations, concomitantly reducing frequency tuning and sensitivity at the CF.

Rather different results have been reported for the effects of quinine on an in vitro preparation of the guinea pig temporal bone: "quinine increased the vibration amplitude at the peak of the mechanical resonance curves and increased the sharpness of tuning."³⁶ These puzzling results, as well as those obtained in the same preparation after acoustic trauma²⁰ (see section on *Effects of acoustic overstimulation*), may imply that the isolated temporal bone preparation has not yet been sufficiently refined as a model of normal cochlear function.

Effects of Systemically Injected Furosemide

Intravenous injection of furosemide, a "loop inhibiting" diuretic, causes a drastic but reversible disruption of cochlear function.⁴ The decisive event is, almost certainly, a precipitous reduction of the normally large (80 mV) positive endocochlear potential, as a result of disruption of metabolically driven ionic pumps in the stria vascularis. Because the drive for the transduction current in hair cells, the apical transmembrane voltage, is substantially (about 50%) determined by the endocochlear potential, furosemide secondarily

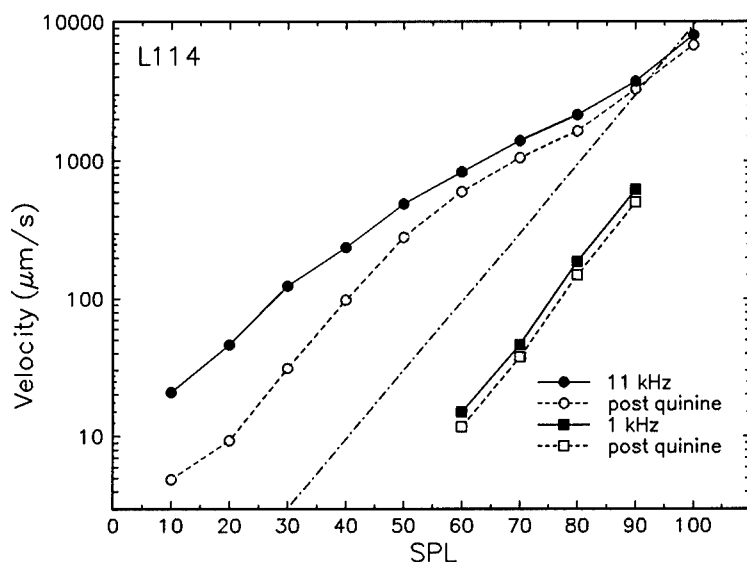


Figure 2-9 Effect of quinine on basilar-membrane responses to tones. Velocity-intensity curves for CF (11 kHz, circles) and 1 kHz tones (squares) were measured immediately preceding and following a 50 mg/kg intravenous injection of quinine. (From Recio and Ruggero.³⁵)

causes a reduction in the hair cell receptor potentials. In turn, to the extent that outer hair cells influence basilar-membrane mechanics, furosemide injection should degrade basilar-membrane responses to sound, diminishing their sensitivity, frequency selectivity, and

intensity-dependent nonlinearity. As shown in Figures 2-10 and 2-11, this expectation is correct.

The mechanical effect of furosemide is both CF specific and intensity dependent.³⁷ Figure 2-10 compares basilar-membrane input-

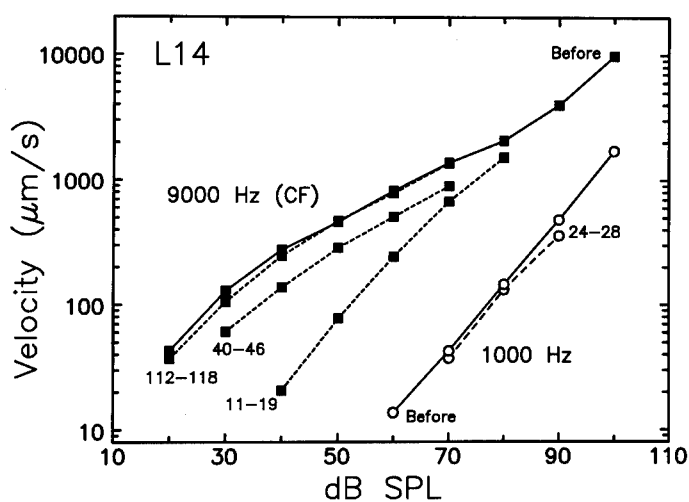


Figure 2-10 Reversible effect of furosemide on basilar-membrane vibration. Input-output curves for responses to tones were recorded (—) immediately preceding and (---) following an intravenous 50 mg furosemide injection. Responses are shown for a CF tone (9 kHz, ■) and for a 1 kHz tone (○). The time of data collection, in minutes relative to the furosemide injection, is indicated for each curve. Reproduced, with permission, from figure 2 of Ruggero and Rich.³⁷

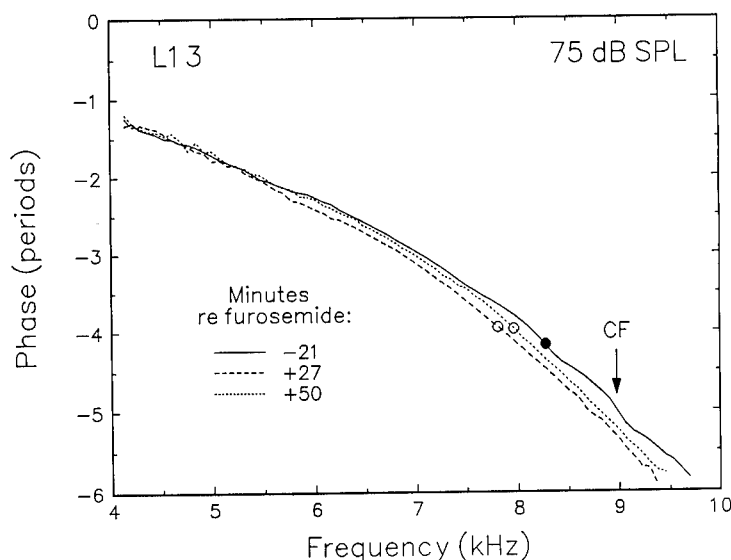


Figure 2-11 Effect of a systemic injection of furosemide upon the phases of chinchilla basilar-membrane responses to clicks. Phases were computed via Fourier transformation of the time averaged responses to 75 dB (peak SPL) clicks. The down-pointing arrow indicates the characteristic frequency (i.e., the frequency of the maximal response at low stimulus levels). The circles indicate the frequencies at which responses to 75 dB clicks were largest before and after furosemide injection. Negative phase values (ordinate, expressed in periods) indicate lags relative to the start of data collection that preceded the arrival of the acoustic click at the eardrum by 250 microseconds. Reproduced, with permission, from figure 6 of Ruggero.³⁸

output functions, measured at various pre- and postinjection times, for tone bursts at CF (9 kHz) and at a frequency much lower than CF (1 kHz). Before furosemide injection, the input-output function at CF consisted of three segments, according to its slope and the stimulus intensity. At low (<30 dB) and high (>90 dB) stimulus intensities, its slope was approximately linear (1 dB of response growth per 1 dB of increase in stimulus level); at intermediate intensities its slope was much shallower, averaging 0.5 dB/dB. For low- and moderate-intensity CF tone bursts, furosemide caused an immediate response reduction as large as 25 dB; at higher stimulus intensities the response reduction was much smaller. In contrast with the strong linearizing effects on the (initially nonlinear) CF responses, furosemide had no apparent effect on the linear input-output function for responses to 1 kHz tone bursts.

Figure 2-11 shows the phase versus frequency characteristics of basilar-membrane

responses to sound before and after injection of furosemide.^{37,38} The phase curves, computed by Fourier transformation of responses to 75 dB clicks, consist of monotonically increasing lags as a function of increasing frequency. Although there is no discernible effect of the furosemide injection for spectral frequencies below 5.5 kHz, postinjection phases lag preinjection phases substantially at frequencies near CF. The furosemide-induced phase lag diminished with the passage of time, accompanying the recovery of response magnitude.

Conclusions

Normal basilar-membrane responses to sound are characterized by a triad of features: high sensitivity, sharp frequency tuning, and nonlinearity (reviewed in Ruggero³⁹). These features manifest themselves only for near CF stimulus frequencies and appear to be inextricably interlinked. When initially present

(i.e., in healthy cochleae), all features of the triad are reduced by ototoxic drugs (e.g., furosemide and quinine) or acoustic trauma and are abolished by death. These mechanical effects of cochlear trauma or death are consistent with the idea that the alterations of basilar-membrane vibration causally determine similar CF-specific effects in auditory-nerve responses. The effects of cochlear insults also imply that the organ of Corti and the basilar membrane sustain a feedback relationship: because the basilar membrane itself is largely acellular, the mechanical effects of noxious agents must be mediated by the organ of Corti, most probably the outer hair cells.³⁷ Thus, the *second filter*,⁷ once posited to account for auditory-nerve frequency selectivity, has been rendered unnecessary, being replaced by a *cochlear amplifier*⁴⁰ residing in the organ of Corti and tightly coupling the electrical responses of outer hair cells to basilar-membrane vibrations.

It is noteworthy that death and noxious agents, while drastically reducing the sensitivity and frequency selectivity of basilar-membrane responses, produce either phase lags near CF (Figures 2-6, 2-11) or only relatively minor phase shifts (e.g., Figure 2-4). A combination of reduced frequency selectivity and phase lags is incompatible with linear minimum-phase systems. For such systems, reduced frequency selectivity is invariably accompanied by decreased delays or, equivalently, phase *leads*. Even in the case of the basilar membrane, which vibrates nonlinearly, one might expect that magnitude changes should be qualitatively correlated with phase changes. Such expectation is founded in the positive correlation that exists in normal cochleae between basilar-membrane frequency selectivity and group delay near CF as a function of stimulus intensity (Figures 2-3, 2-4): basilar-membrane responses to high-intensity stimuli (which are broadly tuned) are accompanied by lower near CF group delays than responses to low-intensity stimuli (which are sharply frequency tuned). In fact, however, cochlear injury or death often induces phase lags in basilar-membrane responses to near CF tones. These phase lags,

which are accompanied by decreases in apparent CF, may result from increased elasticity of the cochlear partition.

Acknowledgments

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Chapter 3

Excitotoxicity and Plasticity of IHC-Auditory Nerve Contributes to Both Temporary and Permanent Threshold Shift

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One area of fundamental and clinical importance is the relationship between the temporary (TTS) and permanent (PTS) threshold shifts induced by acoustic overstimulation. We previously demonstrated in guinea pigs that a continuous pure tone at 6 kHz presented for 15 minutes at 95 dB sound pressure level (SPL), reversibly affects the active mechanisms responsible for the generation of otoacoustic emissions.¹ No obvious ultrastructural abnormality could be seen and a full recovery of the physiological responses was observed 24 hours later. In contrast, when the animals were exposed to more intense sound (6 kHz, 130 dB SPL, 15 minutes), two classical types of damage were found: a pattern of hair cell degeneration in the first row of the outer hair cells (OHCs), then in the inner hair cells (IHCs), and subsequently in the second and the third row of OHCs²; and a massive destruction of dendrites of the primary auditory neurons below the IHCs.³⁻⁶ In addition, it is interesting to note that after 14 days no dendritic damage could be observed, suggesting that a reconnection of the IHCs by the dendrites of the auditory neurons had occurred. In this chapter we examine the acute and long-term effect of an intense sound exposure to determine which abnormalities (hair cell and/or neuronal damage) are responsible for the presence of TTS and/or PTS.

Acute Effects of Acoustic Trauma

When animals were tested 20 minutes after intense sound exposure (pure tone at 6 kHz for 15 minutes at an intensity of 130 dB SPL), the auditory brain stem responses (ABRs) showed a hearing loss greater than 60 dB (Figure 3-1). Histological examination of these cochleas performed in the traumatized area of the basal turn revealed that all dendritic terminals below the IHCs were swollen and disrupted, resulting in an indentation of the IHC basal pole (Figure 3-2a). The OHCs of the first row were drastically damaged (i.e., swollen nucleus and vacuolized cytoplasm; Figure 3-2b). A particularly striking observation is the exceedingly high density of the synaptic vesicles in the medial efferent terminals, whereas the spiral afferent endings look normal, suggesting a very intense metabolic and/or functional activity in these efferents. Altogether, these results suggest that the acute threshold shift is due to neural damage at the dendritic level below the IHCs, and to mechanical damage at the OHC level.

Effects of Acoustic Trauma on Primary Auditory Dendrites

In the central nervous system, it has been shown that an excess of excitatory amino acids leads to a prolonged depolarization of ionic

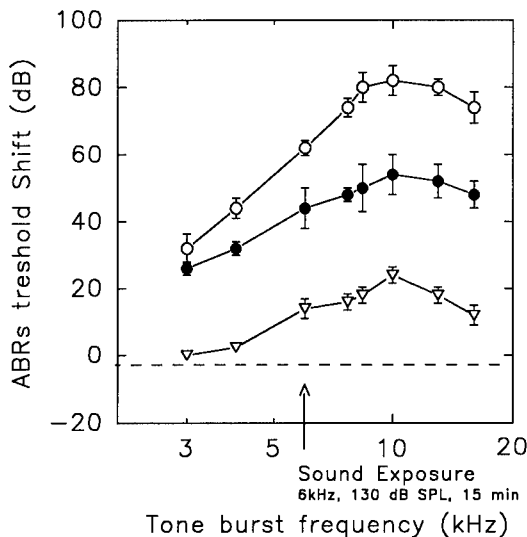


Figure 3-1 ABR threshold shift in decibels (mean \pm SEM) as a function of tone frequency after acoustic trauma. In all animals artificial perilymph was perfused for 10 minutes. This was then followed by a second perfusion for 35 minutes consisting of artificial perilymph alone or containing 5 mM kynureate. Starting 10 minutes after the beginning of the second perfusion period a 6 kHz, 130 dB SPL continuous tone was presented for 15 minutes to the ipsilateral ear. In all animals after the second perfusion period, a third perfusion was finally carried out with artificial perilymph. The threshold shift was defined as the difference between the threshold recorded after the first perfusion with artificial perilymph and the threshold recorded after the third perfusion. The data shown were obtained after intense sound exposure during (○) perfusion of artificial perilymph ($n = 3$), (●) perfusion of artificial perilymph containing 5 mM kynureate ($n = 3$), and (△) 14 days after acoustic trauma ($n = 3$). The data are expressed at mean \pm SEM.

channel-gated postsynaptic receptors, including large cation influxes (Na^+ , K^+ , Ca^{2+}) and a passive entry of Cl^- . The resulting osmotic imbalance then causes a massive water influx into the postsynaptic element, leading to an acute swelling followed by cell death.^{7,8} In the cochlea, the IHCs likely use an excitatory amino acid, probably glutamate, as a neurotransmitter⁹ and excitotoxic mechanisms have

been formally described. Dendritic damage below the IHCs (i.e., a dendritic swelling followed by membrane disruptions) and later a loss of the primary auditory neurons has been reported after *in vivo* perfusion of glutamate agonists.^{10–14} Because similar damage occurred after acoustic trauma, one can speculate that an excess of the release and/or a dysfunction of the uptake mechanism of extracellular glutamate is involved in this phenomena.⁶ Indeed, when 5 mM kynureate (a spectrum glutamate antagonist that postsynaptically blocks endogenous neurotransmitter¹⁵) was applied during the 130 dB SPL sound exposure, no dendritic damage was observed while hair cell damage still persisted. Consistent with these morphological data, the ABR threshold shift was significantly less (about 20 dB between 6 and 16 kHz) than those observed in animals exposed to intense sound during perfusion with artificial perilymph (Figure 3-1). Altogether, these results demonstrate that dendritic damage resulting from acoustic trauma is linked to glutamate excitotoxicity.

Protective Effects of Lateral Efferents During Acoustic Trauma

The physiological activity of the dendritic terminals connected to the IHCs is modulated by the lateral efferent innervation coming from the lateral superior olive. Immunocytochemical studies reported the presence of neuroactive substances such as acetylcholine, GABA, calcitonin gene-related peptide, and certain opioid peptides such as enkephalins and dynorphins in the cochlea.⁹ Although the functional role of these substances needs to be clarified, we recently showed that intracochlear perfusions of 0.01–1 mM of piribédil, a D2 dopaminergic agonist, caused a dose-dependent reduction of the amplitude of the compound action potential (Figure 3-3). This suggests a modulatory action of dopamine via D2 receptors upon the dendrites of the primary auditory neurons. Because this inhibitory action was predominant at high-intensity sound stimulation, one hypothesis is that do-

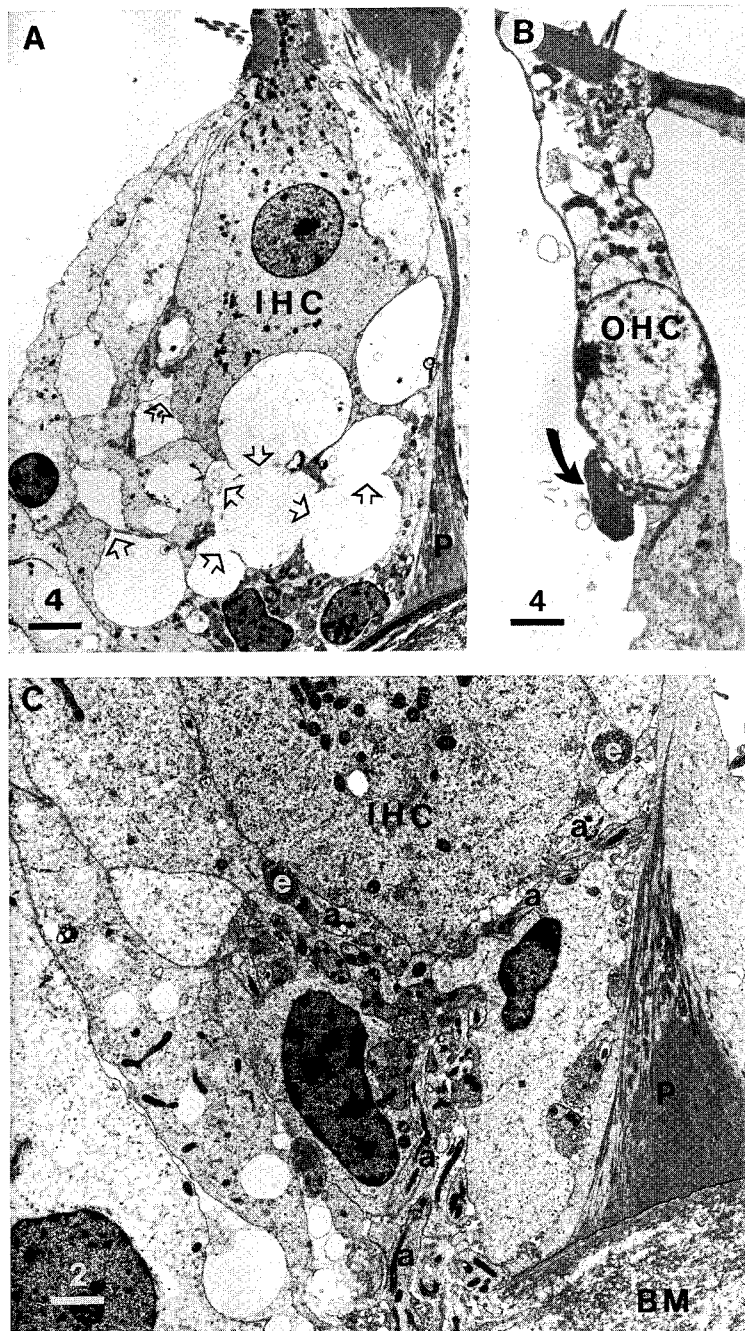


Figure 3-2 Transmission electron microscopy of sensory cells after sound exposure. (a) Inner hair cell (IHC) 20 minutes after intense sound exposure. Fixation took place during the artificial perilymph perfusion (bar 4 μ m). The acoustic trauma induced a dramatic destruction of primary auditory dendrites that resulted in an indentation of the IHC basal pole. This massive swelling of the dendrites underneath the IHCs was followed by a total membrane disruption (arrows). (b) Low magnification of an outer hair cell (OHC) 20 minutes after intense sound exposure. Fixation took place after artificial perilymph perfusion (bar 4 μ m). This damaged OHC comes from the first row of the basal turn. Note the swollen nucleus and the vacuolized cytoplasm. The curved arrow at the base of the cell indicates a dark efferent terminal. (c) Base of an IHC from a cochlea perfused with artificial perilymph containing 1mM picrotoxin (bar 2 μ m). No sign of dendritic swelling was found at the IHC base. The (a) afferent and (e) efferent fibers look normal (P, pillar cell; BM, basilar membrane).

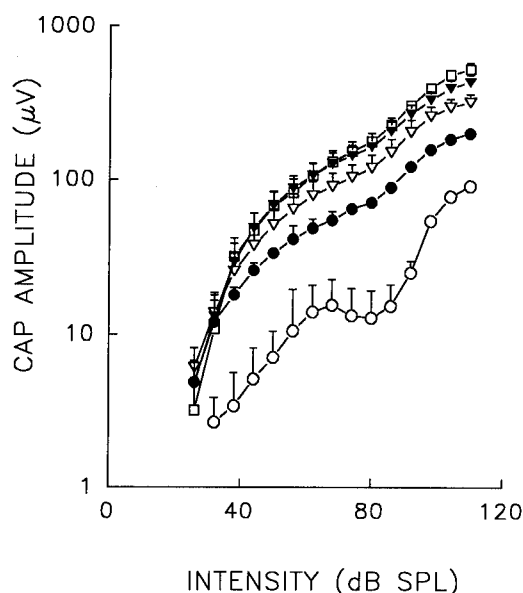


Figure 3-3 Effects of a D2 dopaminergic agonist (piribédil) on the CAP amplitude-intensity function evoked by a 8 kHz tone burst in the guinea pig. When compared with (\square) artificial perilymph, an intracochlear perfusion of (\blacktriangledown) 0.2 mM of piribédil induced no significant changes in the CAP amplitude-intensity function. In contrast, a perfusion of (\triangle) 0.4, (\bullet) 0.8, and (\circ) 1 mM of piribédil caused a dose-dependent reduction in the CAP amplitude, predominantly at the high intensities with only a slight effect at threshold. The data ($n = 8$) are expressed as mean \pm SEM.

pamine could be involved during acoustic trauma, as a lateral efferent transmitter or modulator. Indeed, our present results showed that 1 mM of piribédil applied intracochlearly during an intense sound exposure also prevented the acoustic trauma-induced swelling (Figure 3-2c), except for some fibers contacting the modiolar side of the IHC that were occasionally swollen. This result suggests that the dopaminergic lateral efferents could, to a certain extent, protect the primary auditory neurons via D2 receptors. Although D2 receptors are coupled to the inhibition of adenylate cyclase activity, they also activate K^+ channels.¹⁶ Therefore, one might speculate that the involvement of D2 receptors in the prevention of early damage induced by acoustic trauma might be linked to the activation of K^+

channels to maintain an osmotic balance and thus prevent the acute swelling of the radial afferent endings. Consistent with this assumption is a release of dopamine in rat cochleas submitted to different intensities of noise¹⁷ as has been previously demonstrated for Met-enkephalin,^{18,19} another lateral efferent neuroactive substance.

Long-Term Effects of Acoustic Trauma

Fourteen days after exposure, a partial recovery of the ABR threshold (about 40 dB) was observed (Figure 3-1). Histological examination of these cochleas showed that, while the hair cell damage remained, the base of the IHCs looked normal, suggesting that part of the threshold recovery was due to the reconnection of IHCs by the primary auditory neurons. We have already described such a synaptic plasticity after local perfusion of the glutamate agonist AMPA.²⁰ Therefore, it was interesting to compare this regenerative process with those described after AMPA treatment.

Synaptic Regeneration and Partial Recovery of Threshold After Acoustic Trauma

In both cases (intense sound and AMPA exposure), the time course of regeneration and neosynaptogenesis appears to be relatively fast. One day after acoustic trauma, cochlear neurons have regenerated their dendritic neurites and reached the IHCs to form the first functional synaptic contacts. This observation is supported by the presence of typical synaptic differentiations at contacts between auditory nerve endings and IHCs. At 5 days post-exposure, the pattern of innervation of the IHC looked normal, that is, typical synaptic differentiations were observed at contacts between auditory nerve endings and IHCs. The regenerated dendrites were also normally contacted by efferent endings. In our previous AMPA model, the hair cells remained both morphologically and functionally unaffected while auditory dendrites were completely destroyed. In the present study, intense sound

exposure affected both the primary auditory dendrites and the hair cells. Thus, the neuronal plasticity reported after acoustic trauma does not necessarily account for all of the recovery observed. We should keep in mind that, although kynurenate perfusion reduces the effect of severe acoustic trauma, the threshold shift following kynurenate perfusion was much higher than the shift observed 14 days later (Figure 3-1). Thus, it seems that another repair mechanism, perhaps at the hair cell level, has occurred to partially restore the functional state of the cochlea after acoustic trauma.

Molecular Mechanism Leading to Synaptic Regeneration

In our model of AMPA-induced excitotoxicity, the process of regeneration and neosynaptogenesis was accompanied by strong metabolic reactions within the auditory neurons, such as the expression of glutamate receptors. Previous *in situ* hybridization data indicated that primary auditory neurons expressed mRNAs encoding for GluR2 and GluR3 subunits of AMPA receptors, the NMDAR1 subunit of NMDA receptors, and the mGluR1 subtype of metabotropic receptors.²¹ Consequently, we used a nonradioactive *in situ* hybridization technique to evaluate variations in the expression of mRNAs encoding these receptors. Although no variation in the expression of the GluR2 and the GluR3 mRNAs was seen in primary auditory neurons, the expression of mRNAs encoding NMDAR1 and mGluR1 was increased 1 day after acoustic trauma. This enhanced expression slightly decreased 2 days after sound exposure and returned to a normal value by 3 days, as we previously described in our AMPA model.²⁰ This increased expression of NMDAR1 and mGluR1 mRNAs agrees with data implicating NMDA and metabotropic glutamate receptors in plastic events. A period of transient overexpression of NMDA receptors in different brain regions generally coincides with the period of synaptogenesis and/or the period of experience-dependent synaptic plasticity.²²⁻²⁵ Similarly, a transient

overexpression of metabotropic receptors is involved in postsynaptic protein synthesis in synaptoneuroosomes²⁶ and in the increased inositol phosphate synthesis after kainate-induced epilepsy during the spouting of mossy fibers.²⁷ It is thus tempting to hypothesize that the transient increase in the expression of NMDAR1 and mGluR1 mRNAs in primary auditory neurons reflects an active role of NMDA receptors and inositol phosphate-coupled metabotropic receptors in regeneration and neosynaptogenesis in the cochlea.

Another class of molecules that could play a role in the synaptic plasticity are the neuroactive substances of the lateral efferent system. For instance dopamine, which is one of the putative neurotransmitters of the lateral efferents,⁹ may play an important role. We report herein that dopamine is involved in a protective effect against acoustic trauma-induced excitotoxicity. Moreover, a preliminary finding demonstrated that expression of mRNA for tyrosine hydroxylase is significantly upregulated in the lateral olive neurons 1 day after AMPA-induced excitotoxicity.²⁰ Recent *in situ* hybridization data have shown that mRNAs coding for subtype dopaminergic receptors D1 and D2 are expressed in primary auditory neurons.²⁸ Therefore, we decided to evaluate the variations in the expression of mRNAs encoding these D1 and D2 receptors in the primary auditory neurons. Here again, while the expression of the D1 mRNAs in primary auditory neurons remained unchanged, the expression of mRNAs encoding D2 was increased one day after acoustic trauma (Figure 3-4). Other substances such as enkephalins, of which the release seems to be correlated with noise exposure,^{18,19} may also be involved.

In conclusion, this study is the first demonstration that the primary auditory neurons can regenerate after intense sound exposure and repair their synapses with the IHCs through a mechanism involving NMDA and metabotropic glutamate receptors, as well as those of lateral efferent transmitters. Such a neural regenerative process accounts, at least in part, for the TTS recovery. Considering the relative rapidity of the regenerative process, it is

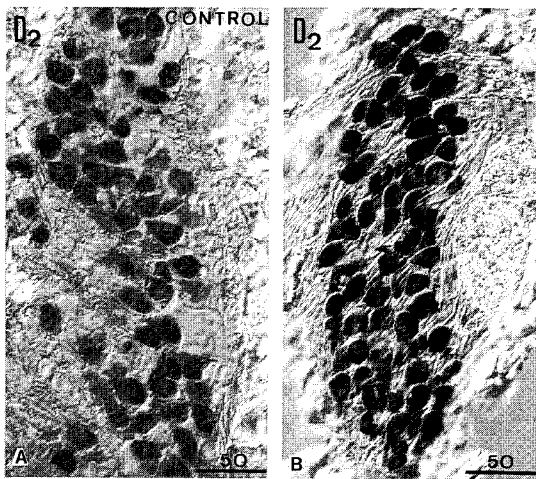


Figure 3-4 Spiral ganglion sections from the basal turn of the guinea pig cochlea. (a) In the animal not exposed to sound, the large number of neurons present in the spiral ganglion expressed mRNAs coding for D2 receptors. (b) In the animal exposed to sound, an increased expression of these mRNAs was observed after 1 day.

tempting to propose that the repair of the synapses could predominate during a first rapid phase (within the 5 days after sound exposure), whereas a second slower phase could depend on another mechanism, perhaps through hair cell repair. Further experiments will be necessary to verify such hypotheses. A last point should be raised concerning a possible neuronal death. Although it is still necessary to determine the number of the primary auditory neurons present after long-term acoustic trauma, some slowly developing neurotoxicity, leading to neuronal death, could also explain the PTS observed herein. A neuronal death occurred in another model of cochlear excitotoxicity in which, 10 days after an intracochlear injection of kainate (which also induces the acute disruption of all the afferent dendrites), 34% of the primary auditory neurons had degenerated.¹¹ In humans, this neuronal death could also occur in different types of cochlear pathologies, especially during such a sensitive period as aging²⁹ that is marked by the frequent occurrence of vascular atrophy in the cochleas.³⁰ Accordingly, a significant loss of primary auditory neurons was actually re-

ported in some types of presbycusis (deafness in the elderly) called neural presbycusis.³¹

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Chapter 4

Noise-Induced Expression of Heat Shock Proteins in the Cochlea

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Heat Shock Responses

Exposure to heat induces a physiological response called the *heat shock response* in all cells and organisms. The phenomenon was first described by Ritossa¹ who observed chromosomal puffing in *Drosophila* after exposure to heat. This heat-induced chromosomal puffing was associated with the synthesis of specific proteins called heat shock proteins (HSPs).² Although first described after hyperthermic stimuli, HSPs are now commonly referred to as stress response proteins because of their expression after a variety of other stress including nutrient deprivation, oxygen radicals, viruses, ischemia, and xenobiotics, etc. as well as heat shock. The role of HSPs has been studied and it is now clear that they have a function in unstressed as well as stressed cells.³ In unstressed cells HSPs play a role in chaperoning other proteins and aid in transmembrane transport of proteins. They also have functions involving facilitating proper folding of proteins as well as preventing improper folding and protein aggregation. In these roles HSPs are thought to protect cells from the deleterious effects of various cellular stresses that promote misfolding or protein aggregation.^{4,5} The HSPs are usually classified and named according to their apparent molecular weight. The 70 kDa HSP family includes the inducible form (72 kDa) synthesized with stress, and constitutive forms (70, 73 kDa) normally present in unstressed cells as well as 78 kDa glucose-regulated protein. The low

molecular weight HSPs belong to the 20 kDa family; the high molecular weight HSPs (83, 90, 110 kDa) and the 94 kDa glucose-regulated protein belong to the HSP 90 family. Different HSPs have been proposed to act through different mechanisms: HSP 27 through influencing actin depolymerization,⁶ HSP 60 influencing mitochondria as molecular chaperones, HSP 72 through protein folding and chaperoning, and HSP 90 influencing steroid related mechanisms.^{7,8}

Heat Shock Protein in Auditory System

A number of environmental and physiological stresses, for example, noise, ototoxic drugs, and hypoxia, are known to cause transient or permanent pathological damage to cochlea (for review see Lim⁹ and Libermann¹⁰). The auditory system often has the capacity to rebound from dysfunction induced by environmental stresses, termed temporary threshold shifts (TTS). Suggestions for the cause of TTS range from disruption of the active biomechanical process to loss of stereocilia stiffness. The mechanisms behind TTS, however, have yet to be elucidated and we do not understand how and why protective mechanisms may fail. HSPs are believed to be induced by moderate stresses in order to protect the cells from even more severe stresses.¹¹ It can be hypothesized that HSPs may have some responsibilities in protective mechanisms of hair cells against various kinds of insults. Several questions have been raised to

examine the possible roles of HSPs in the auditory system. Are stress shock proteins expressed in the cochlea? What stress might induce them? In what cells are they expressed? Several investigators have demonstrated the induction of HSPs in guinea pig and rat cochlea. In normal unstressed guinea pigs there was a constitutive level of expression of HSP 72 in Deiters cells and interdental cells of spiral limbus¹² and the expression was increased with hyperthermic stress.¹³ On the other hand there was no expression of HSP 72 in unstressed rat cochlea and its expression was induced by heat¹⁴ and transient hypoxia.¹⁵ Recently we have shown expression of HSPs in rat cochlea after noise overstimulation.^{16–18}

Stress Response in Cochlea With Noise Overstimulation

Although the expression of HSP was induced in cochlea with heat and ischemic stress, some limitations remain to explain the role of HSPs in the cochlea because the stresses were not so specific and physiologic to the cochlea. We have examined if they may have a general protective role in the auditory system as well as a specific role in protection from noise-induced hearing loss. Three different kinds of HSPs (HSP 72, HSP 90, and HSP 27) were evaluated in this series of studies.

Induction of HSP 72 With Noise Exposure

The HSP 70 family is the most conserved and best studied among HSP families. In mammalian cells there are two major members of the 70 kDa family: an abundant, constitutive 73 kDa protein and a highly stress inducible 72 kDa protein.¹⁹ Immunocytochemical and Western blot analyses were used to detect the HSP 72 expression in rat cochlea after noise exposure. Sprague–Dawley rats were used in this study. Experimental animals were exposed to 110 dB broad band noise with pulses (5/second 50% duty cycle) for 1.5 hours in a sound proof booth. The control animals did not receive any stimulation. Four, 6, and

8 hours after stimulation, subjects were sacrificed and cochleae were microdissected or decalcified with 3% EDTA for cryostat section. The cochlear tissues were processed for immunochemical detection of HSP 72 or Western blot analysis. Immunocytochemical results showed HSP 72 expression after noise exposure compared to no expression in control, non-noise exposed specimens (Figure 4-1). HSP 72 immunoreactive staining was observed in all three rows of outer hair cells (Figure 4-2a) and stria vascularis (Figure 4-2c). Maximal immunoreactive staining was observed 6 hours after stimulation. Immunostaining was mainly in the cytoplasm of outer hair cells without nuclear staining (Figure

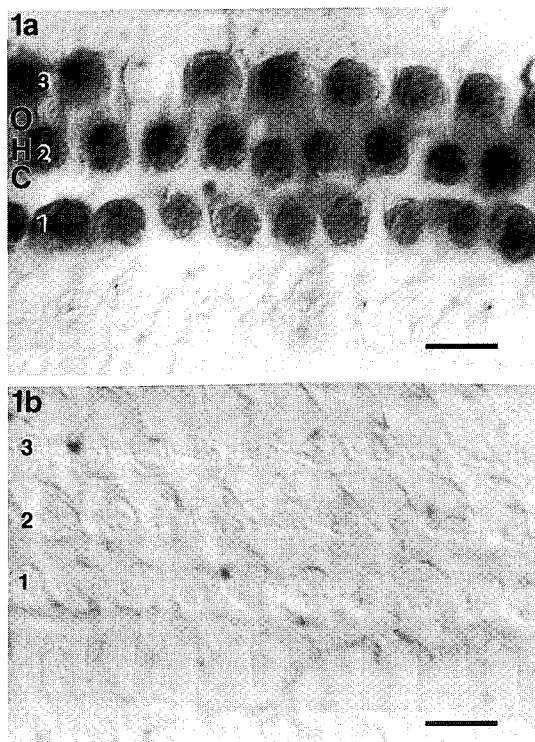


Figure 4-1(a) Surface preparation from the second turn of rat cochlea stimulated with 110 dB SPL of broad band noise immunoreacted with antibody to HSP 72. HSP 72 immunoreactivity is seen in all three rows (1–3) of outer hair cells. **(b)** Surface preparation from the second turn of nonstimulated rat cochlea immunoreacted with antibody to HSP 72. No HSP 72 immunoreactive staining is seen. OHC, outer hair cells; scale bar = 10 μ m.

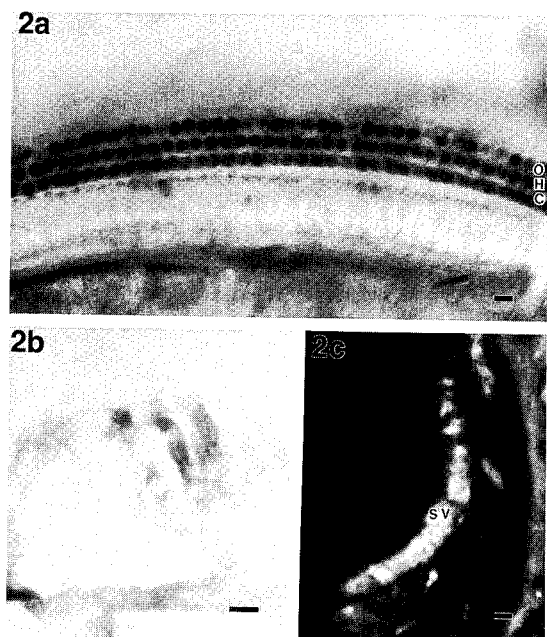


Figure 4-2 Surface preparations and sections of rat cochlea stimulated with 110 dB SPL of broad band noise, sacrificed 6 hours after noise exposure and immunoreacted with antibody to HSP 72. (a) Surface preparation from the second turn of rat cochlea with immunoperoxidase staining. Three rows of outer hair cells are intensely immunostained for HSP 72 and a few immunostained inner hair cells are also seen. OHC, outer hair cell; scale bar = 10 μ m. (b) Cryostat section from the second turn of rat cochlea with immunoperoxidase staining. HSP 72 immunostaining of outer hair cells is seen, mainly in their cytoplasm, without nuclear staining. Scale bar = 10 μ m. (c) Cryostat section from the second turn of rat cochlea with immunofluorescent staining. Intense HSP 72 immunoreactive labeling is seen in the stria vascularis. SV, stria vascularis; scale bar = 10 μ m.

4-2b). Only a few immunoreactive stained inner hair cells were seen and supporting cells and spiral ganglion cells were not stained. Western blot results also show HSP 72 synthesis with noise exposure. Intense bands were found at 72 kDa molecular weight (MW) in tissues from sensoriepithelium with the lateral walls of the cochlea and modiolus, including the spiral ganglion and auditory nerve of noise-exposed rat cochlea. In control rats very light bands, much weaker than in noise-

exposed rats, were seen. HSP 72 expression with hyperthermic stress was seen in spiral ganglion cells¹⁴ and interdental cells of the spiral limbus,¹³ and noise-induced expression of HSP 72 was seen in rat outer hair cells. These immunocytochemical results are comparable to the previous results of our laboratory¹⁶ that showed HSP 72 expression in outer hair cells following transient ischemia. It can be presumed that hypoxia and acoustic overstimulation are more effective than hyperthermia at inducing HSP 72 in outer hair cells. The detection of HSP 72 immunostaining in outer hair cells but not inner hair cells may be due to several factors. Outer hair cells may be more stressed than inner hair cells and therefore more likely to express HSP 72; or outer hair cells may have a lower threshold for induction of HSP 72. Further study will be necessary to examine if inner hair cells express HSP 72 under specific stimulus conditions for inner hair cells.

Expression of HSP 72 With Noise-Induced TTS

To evaluate the functional roles of HSP 72 in the cochlea after noise exposure, we examined the relationship between HSP 72 expression and threshold shift and scar formation in rat cochlea. Auditory brain stem responses (ABR) were used to check threshold shift before and after noise exposure. Actin-specific phalloidine and anticytokeratin antibody were used to evaluate scar formation in the organ of Corti after noise exposure. No hair cell loss, swelling of supporting cells, or other signs of scar formation were observed in the organ of Corti 6, 12, and 24 hours after stimulation (Figure 4-3a,b). No scar formation and hair cell loss with the presence of HSP 72 expression after acoustic overstimulation suggests that rat cochlea might be more resistant to intense noise exposure than guinea pigs who showed scar formation 6 hours after noise overstimulation.²⁰ Or noise-induced HSP 72 synthesis might protect hair cells from further damage.

Testing of ABR showed 30–50 dB threshold shifts lasting 3–4 hours after noise exposure.

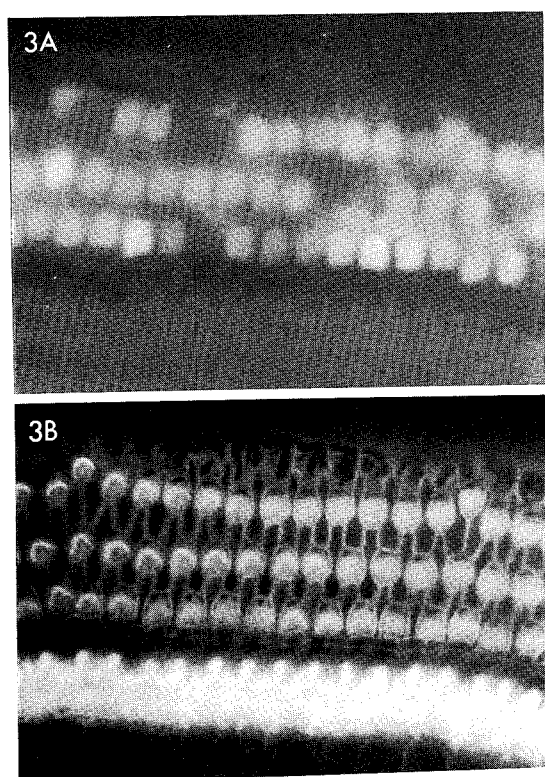


Figure 4-3(a) Surface preparation of rat cochlea stimulated with 110 dB SPL of broad band noise, sacrificed 24 hours after noise exposure and immunoreacted with antibody to HSP 72 and rhodimine conjugated secondary antibody. Immunostained outer hair cells are seen. **(b)** Same tissue as **(a)** stained with FITC conjugated actin specific phalloidine. No hair cell loss, swelling of supporting cells, or other signs of scar formation are seen.

One animal did not show temporary threshold shift, even though the sound exposure was the same as the others. In this animal no expression of HSP 72 was seen. This result suggests that HSP 72 is induced under noise exposure conditions that lead to temporary threshold shift. The role of HSP 72 remains to be determined. If HSP 72 has a protective function related to TTS, it is more likely to be involved in restoration of function than in an initial protection, based on its time course of expression, peaking at around 6 hours after stimulation. On the other hand HSP 72 may have a protective function that is completely

unrelated to the events underlying TTS or not correlated with its peak of expression.

Noise-Induced Increased Expression of HSP 90

Although HSP 72 is the most conserved and best studied among the HSP family, another high molecular weight heat shock protein, HSP 90 may also have an important role in cellular protection against stresses. Mammalian HSP 90 is a very abundant protein in cells grown under normal conditions and its synthesis increases after heat shock treatment.²¹ HSP 90 has been suggested to be involved in steroid-related mechanisms and may also have an involvement with actin.²² We investigated the expression of HSP 90 in normal cochlea and whether noise exposure could increase the synthesis of HSP 90. Immunocytochemical analysis were used to detect HSP 90 expression in the cochlea using anti HSP 90 monoclonal antibody (StressGen). Immunoreactive staining was viewed in surface preparations of the cochlear spiral. The level of noise exposure utilized resulted in a temporary threshold shift (30–40 dB) for 3–4 hours. A constitutive (noninduced) level of HSP 90 immunoreactivity was seen in the non-noise-exposed normal animals, in both inner and outer hair cells. An increased intensity in HSP 90 immunoreactive staining of inner and outer hair cells was seen 5 and 6 hours after noise overstimulation (Figure 4-4). Noise-induced expression of HSP 90 is different from the noise-induced expression of HSP 72. First, for the HSP 70 family there are separate constitutive and induced proteins, so that no constitutive (noninduced) HSP immunoreactive staining was observed. HSP 90, on the other hand, showed constitutive staining in the cochlea, as it does in the other regions. Second, HSP 72 immunoreactive staining was only seen in outer hair cells. HSP 90 immunoreactive staining, both constitutive and induced, is seen in both inner and outer hair cells. The increased expression of HSP 90 after noise overstimulation may have a role in the protection of hair cells from acoustic overstimu-

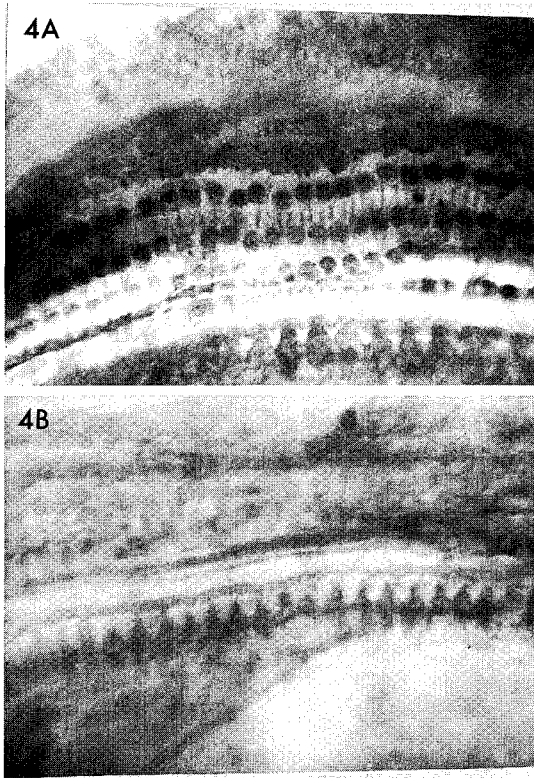


Figure 4-4 Surface preparations of rat cochlea stimulated with 110 dB SPL of broad band noise, sacrificed 5 hours after noise exposure and immunoreacted with antibody to HSP 90. Immunoreactive staining was seen in (a) outer hair cells and (b) inner hair cells.

lation. However, the timing of expression compared to the TTS also suggests a role in recovery. We also saw a noise-induced decrease in HSP 90 immunoreactivity. It is possible that there is an initial decrease in HSP 90 levels, as constitutive HSP 90 is utilized, and then a large increase as the induced HSP 90 is being expressed. We need to further investigate the temporal dynamics of HSP 90 response. HSP 90 is believed to complex with many steroid hormone receptors including the stress-related glucocorticoid receptors. It will be interesting to determine the relationship between HSP 90 and steroid receptors in the cochlea and the functional roles of HSP 90 in the cochlea as well.

Noise-Induced Expression of HSP 27

A lower molecular weight stress protein, HSP 27, was originally identified in smooth muscle and is suggested to have a possible role in actin depolymerization and stabilization.⁶ It is also known to protect actin during hyperthermia or exposure to cytochalasin D *in vitro*.²³ Because noise overstimulation appears to affect the polymerization of actin in hair cell stereocilia, HSP 27 might possibly play a role in the cochlea. The presence of HSP 27 with and without noise overstimulation (110 dB BBN for 1.5 hours) was detected by immunoperoxidase immunocytochemistry using monoclonal antibody to HSP 27 (Sigma) on surface preparation of noise-exposed and non-noise-exposed rat cochlea. Nonexposed cochlea showed constitutive HSP 27 immunoreactive staining in stereocilia of inner and outer hair cells and a light cytoplasmic staining of outer hair cells. Noise-exposed animals showed an increased staining of outer hair cells most prominent in the apical half of the cochlea. HSP 27 has a constitutive level in stereocilia (Figure 4-5a) that could be involved in depolymerization of actin and induced level in outer hair cells (Figure 4-5b) that could be involved in repolymerization.

Conclusion

Heat shock proteins are expressed in the cochlea from a variety of stresses. Noise induces expression of HSP 72 and HSP 27 in outer hair cells and modulates the levels of HSP 90. HSP 90 has a constitutive expression in inner and outer hair cells and HSP 27 has constitutive levels in stereocilia. It is interesting to consider the protective role that HSPs could provide against noise-induced hearing loss. Constitutive HSPs could help stabilize proteins against the initial stress caused by noise and induced HSPs could provide further stabilization of proteins, renaturation of affected proteins, and stabilization of receptors and cell processing. HSP 27, 72, and 90 have all been reported to bind to and affect certain cytoskeletal proteins including actin and spec-

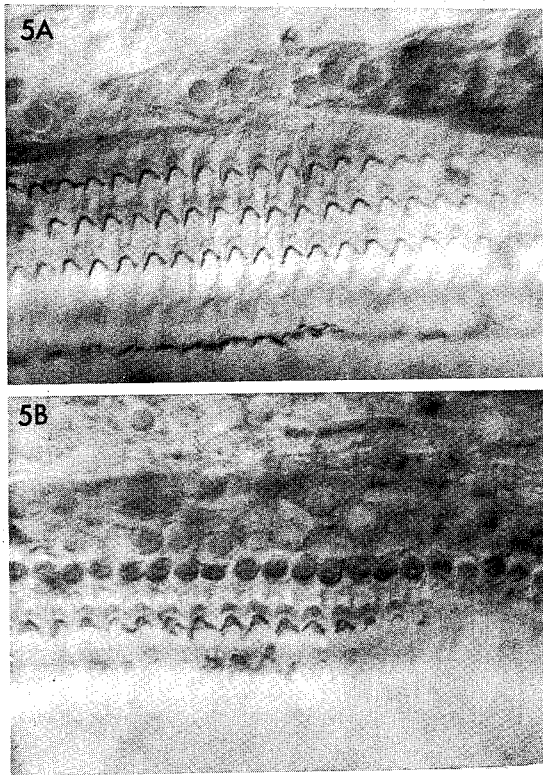


Figure 4-5(a) Surface preparations of normal non-noise-exposed rat cochlea. HSP 27 immunoreactive staining is seen on the stereocilia of outer hair cells. **(b)** Surface preparations of rat cochlea stimulated with 110 dB SPL of broad band noise, sacrificed 5 hours after noise exposure and immunoreacted with antibody to HSP 27. Intense HSP 27 immunoreactive staining is seen in the outer hair cells.

trin, that have important roles in hair cell functioning. HSP 27, in particular, is closely related to actin. These could influence stereocilia rigidity and/or linkages that may be related to TTS and its recovery. HSPs have also been related to acquired tolerance to stresses in many systems and they may be involved in toughening/conditioning in the cochlea (see chapters in this volume by Henderson et al. and Canlon and Dagli). Further studies will be necessary to determine the role(s) of HSPs in the cochlea, how they achieve their function, and how they may work in protective mechanisms against noise damage as well as other stresses.

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Chapter 5

Changes in Gene Expression Following Temporary Noise-Induced Threshold Shift

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Many hypotheses have been advanced to explain temporary threshold shift (TTS) due to intense sound exposure including, among others, changes in the ionic composition of the cochlear fluids,^{1,2} cochlear blood flow,³ stereociliary morphology,^{4,5} hair cell metabolism,⁶ or motility,⁷ and the synapses between hair cells and spiral ganglion neurons.⁸ Most such hypotheses are based upon disturbances of the cochlear transduction process or of cochlear homeostasis. Definitive evidence to confirm any of these potential etiologies is lacking, and the causes of TTS remain unknown. However, it is quite possible that TTS involves changes in proteins that are important for some aspect of the cochlear transduction process or of cochlear homeostasis. Many of these proteins have been identified in the cochlea and are available for study, but not all are known.

The proteins that subserve important functions in a cell are typically encoded by mRNA in that cell. Only proteins that originate at other sites or are subject to very low turnover rates, will not be so represented. Given this fact, it is possible to identify many of the proteins present in cochlear cells based on their mRNA sequences. We have employed *in situ* mRNA hybridization and/or polymerase chain reaction (PCR) to document the inner ear expression of a number of genes encoding proteins either known or suspected to play a role in cochlear function.^{9–11} One such molecule is Na,K-ATPase, which is abundant in several cochlear tissues and appears to be the

primary ion transporting enzyme responsible for the generation of endolymph and the endocochlear potential.^{12,13}

Na,K-ATPase consists of two subunits. The α subunit is a polypeptide of molecular weight (MW) 100 000 containing both the catalytic site for ATP hydrolysis and the ion exchange site. The function of the β subunit, a glycoprotein of MW 55 000 that exposes most of its mass to the extracellular side of the cell membrane,¹⁴ is less certain. Three molecular isoforms of the α subunit and two isoforms of the β subunit have recently been identified, each encoded by a separate gene located on different chromosomes.^{15,16} Northern- and Western-blot analyses have identified distinct isoform distributions and prevalences in different tissues and developmental stages. In the adult rat $\alpha 1$ is expressed at the highest levels in the kidney and heart; lower levels are detected in the brain, lung, spleen, and lactating mammary gland.¹⁷ $\alpha 2$ is strongly expressed in the brain and at lower levels in the heart and lung.¹⁸ Expression of $\alpha 3$ is restricted entirely to the brain.^{17–19} The $\beta 2$ subunit is abundant only in brain, while $\beta 1$ is predominant in kidney and heart and is detected at lower levels in lung.^{16,17,19} This variation in tissue distribution suggests a functional subspecialization among the different possible combinations of Na,K-ATPases. For example, $\alpha 1$ and $\beta 1$ are the prevailing isoforms in the adult kidney; the pineal gland contains primarily $\alpha 3$ and $\beta 2$ isoforms.¹⁸ Different isoform combinations appear to have different substrate

affinities,^{14,18} and thus would perform most efficiently under different physiological conditions.

All five subunit isoforms are present in the inner ear, with different combinations found in different tissues.⁹ For example, in the stria vascularis the only isoforms are $\alpha 1$ and $\beta 2$. The purpose of the present study was to determine whether the levels of mRNAs encoding Na,K-ATPase were altered following exposure to noise of sufficient intensity to produce a TTS.

Methods

Subjects

Young adult (60–90 days old) Sprague–Dawley rats were used for subjects in all studies.

Auditory Brain Stem Response (ABR)

Rats were anesthetized with ketamine (50 mg/kg), rompun (20 mg/kg), and acepromazine (10 mg/kg). A needle electrode was placed on the vertex, a coil electrode on the roof of the mouth, and a reference needle electrode in the neck musculature. ABR thresholds were measured at 1, 2, 4, 8, 16, and 32 kHz using an HP 3561A dynamic frequency analyzer and a descending method of limits. See Ryan et al.²⁰ for additional details. Thresholds were measured prior to noise exposure, as soon as possible after noise exposure, and at least 2 weeks later.

Noise Exposure

White noise was bandpass filtered at 1414–5656 Hz (Rockland 852), amplified (Crown D-90), attenuated (HP 350D), and applied to a JBL 2482 midrange speaker coupled to an Altec 511B horn. Intensity was set to 110 dB SPL using a Bruel and Kjaer 2209 noise level meter and 1613 octave band filter, by averaging the readings obtained in the 2 and 4 kHz bands. Rats were placed in a 15 × 30 cm wire screen cage suspended in a double-walled sound-attenuated room (IAC 1200A) with an anechoic lining 15 cm from the horn aperture,

and exposed for 1 hour. Details of the noise stimulus have been presented elsewhere.²⁰

In Situ Hybridization

Rats were deeply anesthetized with nembutal (50 mg/kg) and perfused intravascularly with 50 mL of warm saline followed by 150 mL of 4% paraformaldehyde (pH 6.5) and then 150 mL of 4% paraformaldehyde plus 0.1% glutaraldehyde (pH 9.5). The inner ears, including the cochlea, vestibular labyrinth, and endolymphatic sac, were postfixed at 4°C overnight in the final perfusate, sunk in 4% paraformaldehyde (pH 9.5) plus 30% sucrose, frozen in OCT, and sectioned on an AO cryostat at 20 μ m.

The procedures used for in situ hybridization were as described in detail elsewhere.^{21,22} Briefly, tissue sections were permeabilized with 0.0002% proteinase K for 30 minutes at 37°C, then hybridized to ³⁵S-UTP labeled antisense riboprobes that were synthesized from cDNA templates of genes coding for three isoforms of the α subunit and two isoforms of the β subunit of rat Na,K-ATPase (see Ryan and Watts⁹ for details of probe sizes and placement within the mRNAs). The corresponding sense strand riboprobes were synthesized and served as negative controls for each probe. After hybridization at 56°C overnight, the sections were then treated with ribonuclease A and high stringency washes (low salt and high temperatures). Preliminary evaluation of hybridization was obtained by opposition of slides to X-ray film for 48–72 hours. Afterward the slides were coated with Kodak NTB-2 liquid autoradiographic emulsion, exposed at 4°C for 2–4 weeks depending on the strength of the signal obtained on the film, developed in Kodak D-19 (2.5 min at 14°C), and fixed in Kodak rapid fixer. The sections were counterstained through the emulsion with the nuclear stain bisbenzamide (0.001% for 2 minutes). The sections were examined by fluorescence microscopy to identify tissue structures and cells, and under dark field microscopy to evaluate the distribution of autoradiographic grains.

Quantitative analysis was performed on film images from sections that had been processed together using the same probe, exposure, and development. Optical density was measured in a constant area within the image corresponding to the stria vascularis for each cochlear turn, and averaged. Film background was measured from adjacent areas not opposed to tissue, averaged, and subtracted from the stria vascularis value. Because optical densities are ratios, the data were analyzed using the Wilcoxon signed rank test.

Results

Noise-Induced Hearing Loss

The threshold shift measured immediately after exposure to the 110 dB SPL stimulus is shown in Figure 5-1. The loss was greatest at 4 kHz, where it averaged 22.5 dB. By 2 weeks after exposure, thresholds had recovered to preexposure levels, demonstrating that the noise-induced loss consisted entirely of TTS.

In Situ Hybridization

The patterns of expression of Na,K-ATPase mRNA in control cochleas was identical to those described previously.⁹ In particular, both $\alpha 1$ and $\beta 2$ mRNAs were strongly ex-

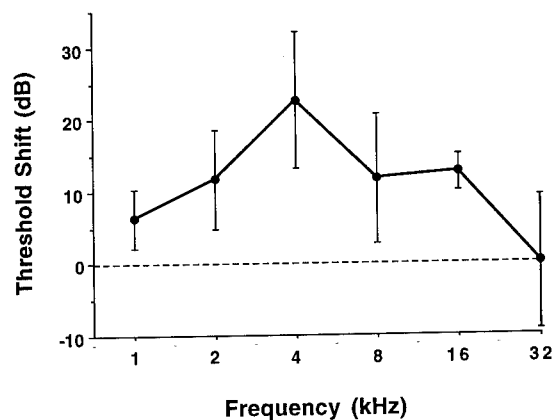


Figure 5-1 Average ABR threshold shift measured in three rats immediately after exposure to a two-octave (1.4–5.6 kHz) band of noise at 110 dB SPL for 1 hour. Vertical bars represent 1 standard deviation above and below each mean.

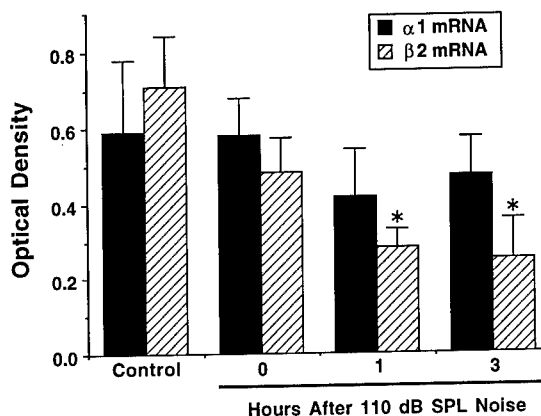


Figure 5-2 Hybridization of riboprobes complementary to mRNA encoding the $\alpha 1$ and $\beta 2$ subunits of the Na,K-ATPase in the stria vascularis of normal rats and of noise-exposed rats at various survival times after noise exposure. Note that $\beta 2$ hybridization is significantly reduced 1–3 hours after exposure. Vertical bars represent 1 standard deviation. (*) Significantly different from control ($p < 0.05$, Wilcoxon signed rank test).

pressed in the stria vascularis, and they were the only isoforms present. The combination of $\alpha 1$ and $\beta 1$ mRNA was observed in the spiral ligament; $\alpha 3$ and $\beta 1$ were strongly expressed in spiral ganglion neurons. Very little Na,K-ATPase mRNA was expressed in the organ of Corti. No differences in the cochlear expression of $\alpha 1$, $\alpha 2$, $\alpha 3$, and $\beta 1$ mRNA were apparent after TTS-inducing noise exposure. However, a transient depression of hybridization with the $\beta 2$ probe was observed in the stria vascularis (Figure 5-2). Immediately after noise exposure, $\beta 2$ hybridization was slightly lower than, but not significantly different from, that seen in nonexposed controls. At 1 and 3 hours postexposure, $\beta 2$ expression was significantly lower than in controls ($p < 0.05$). By 24 hours after exposure, the level of $\beta 2$ mRNA had returned to preexposure levels.

Conclusions

In summary, we found that production of mRNA encoding the $\beta 2$ isoform of the β subunit of Na,K-ATPase is transiently reduced in the stria vascularis after a TTS-inducing noise

exposure, but expression of the $\alpha 1$ isoform of the α subunit does not change. This result suggests that the stria vascularis may play a significant role in TTS.

A number of other investigators have noted changes in the stria or in the endolymph following intense noise exposure. For example, Duvall et al.²³ noted reversible changes in strial ultrastructure after a brief exposure to intense noise in the chinchilla. Several investigators have observed changes in the strial vasculature.³ Salt and Konishi¹ and Jian et al.² noted a decline in the guinea pig endocochlear potential (EP). Salt and Konishi¹ and Li et al.²⁴ found that the potassium concentration of endolymph decreased after exposure to intense sound. Our data provide further evidence that noise can influence strial function.

The functional role of the β subunit of ion transport ATPases is not entirely clear. Although it does not contain the sites for hydrolysis of ATP or ion exchange, its presence seems essential for ion transport, because its enzymatic separation from the α subunit results in an irreversible inactivation of ion pump activity.²⁵ Also, yeast cells transformed with DNA encoding only the α subunit produce nonfunctional Na,K-ATPase.²⁶ There is evidence that the β subunit is involved in protein folding of the α subunit, in transport of the mature enzyme from the endoplasmic reticulum, and in its insertion into the cell membrane.^{25,27} The different subunit combinations of Na,K-ATPase exhibit different affinities for substrate.^{14,18} The presence of the $\beta 2$ subunit is associated with transport against a high sodium gradient, and it has been suggested that the $\beta 2$ subunit increases the binding efficiency of the α subunit for sodium.²⁵ As we have noted in the cochlea,⁹ the predominance of $\alpha 1$ and $\beta 2$ in the cells believed to generate endolymph and the endolymphatic potential supports this concept of $\beta 2$ subunit function, because the electrochemical gradient for sodium between endolymph and perilymph is very steep. The presence of some $\beta 1$ mRNA in addition to $\alpha 1$ and $\beta 2$ in vestibular dark cells,²⁸ when the stria vascularis has none, could be related to the lower resting potential and higher sodium content of ves-

tibular endolymph, which would reduce the sodium gradient.

Whether the reduction in $\beta 2$ mRNA over a period of hours reflects a change in enzyme function in stria vascularis is not at all clear. The high level of mRNA production in the normal stria vascularis suggests rapid turnover of the enzyme, and thus reduced $\beta 2$ mRNA could lead to a decrease in insertion of the $\alpha 1$ subunit into the basolateral membrane of the marginal cell. This could in turn lead to reduced enzyme activity. A reduction in enzyme activity, with a subsequent decrease in the EP, could be a means of adaptation to high levels of activity in the organ of Corti. However, it may be that the downregulation of $\beta 2$ mRNA is too brief to be reflected in actual enzyme function, or that it produces only a modest change.

Alternatively, the reduction in $\beta 2$ mRNA may be a response to changes in the endolymph. It seems reasonable to assume that the high level of expression of Na,K-ATPase mRNAs in the stria is under regulatory control, and that it may be influenced by the electrochemical environment of the stria. Changes in the electrochemical environment, such as decreases in EP or changes in endolymph ion concentrations, could interact with the normal regulatory mechanisms of Na,K-ATPase expression to produce the downregulation of the β subunit message. Regulation of Na,K-ATPase mRNA by intra- and extracellular sodium has been documented in other systems.^{29,30} In any event, the results suggest that production of the $\beta 2$ subunit can be regulated by excessive cochlear activation, and that the transcription of the β subunit gene may be a critical point for regulation of Na,K-ATPase in the stria. In support of this conclusion, changes in β subunit mRNA have been shown to regulate Na,K-ATPase in kidney cells.²⁵

Acknowledgments

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Chapter 6

Genetic Susceptibility to Noise-Induced Hearing Loss in Mice

Lawrence C. Erway and James F. Willott

Genetic Speculations Regarding Noise-Induced Hearing Loss

Presently, little is known about how susceptibility to noise-induced hearing loss (NIHL) is affected by genetic variables. Indeed, we are currently at the stage of being able to do little more than speculate about potential avenues by which genes may influence NIHL. Thus, speculation seems like a good point to begin a discussion of possible ways that genes can affect susceptibility of the inner ear to noise damage.

Genes might cause histopathological and/or pathophysiological changes in hair cells or other cochlear structures that render them more or less vulnerable to additional trauma. For example, numerous genetic mutations are known to cause cochlear pathology in mice,¹ and many genetically determined syndromes in humans are associated with cochlear pathology.^{2,3} It is feasible that some of the gene actions that produce cochlear pathology also affect the ear's vulnerability to noise, or that changes in vulnerability are secondary to cochlear pathology. Several inbred mouse strains are known to possess genes that produce cochlear pathology during adulthood.⁴ Most notably, inbred C57BL/6J (C57) mice possess a gene for age-related hearing loss (AHL) that appears to make them more susceptible to NIHL, a topic discussed in detail below.

Certain genes might not cause cochlear pathology per se, but rather produce inadequacy of biochemical or metabolic mechanisms that

normally protect healthy cochlear tissue from acoustic overexposure. For example, different normal hearing strains of mice differ with respect to susceptibility to NIHL,⁵⁻⁷ suggesting an action such as this. Mechanisms of this kind might involve such variables as glucose/oxygen metabolism vis à vis recovery from or resistance to physiological stress associated with intense noise (cf. Proctor³). One possible example is melanin pigment. Although the evidence is not totally consistent,⁸ melanin appears to protect the cochlea from NIHL under certain conditions (see citations in various authors⁸⁻¹¹). For example, albino guinea pigs⁹ or humans with less melanin^{10,11} have been shown to be more susceptible to NIHL.

Genetic actions on nonauditory functions (e.g., autonomic, vascular, endocrine, etc.) might render the ear more or less vulnerable to noise. Several examples may be relevant. Rats genetically predisposed to high blood pressure exhibit greater noise-induced cochlear pathology than normal rats.¹² Human workers with high serum cholesterol (a trait that can be influenced genetically) appear to be slightly more susceptible to damage from industrial noise.¹³ A rather striking example of a genetic circulatory system disorder is sickle cell anemia, a trait that could result in inadequate blood supply to the inner ear.¹⁴ There seems to be a relationship between renal dysfunction and cochlear dysfunction,¹⁵⁻¹⁷ and kidney function is known to be affected by certain genes.^{18,19}

Genes might cause increased vulnerability to nonacoustic ototraumatic factors (e.g., ototoxic drugs) that might interact with the effects of noise exposure. Ototoxicity can potentiate noise-induced damage^{20,21} with marked individual variability in susceptibility to at least some ototoxic drugs,²² suggesting a genetic influence.

If middle ear conductive processes were affected by genes, noise could become more effective. Examples might include variations in the effectiveness of the acoustic reflex (e.g., muscles, innervation) or physical properties of the outer/middle ear (e.g., immittance properties). Middle ear disorders that can be genetically transmitted include otosclerosis and Paget's disease.^{2,3}

Genes can influence a variety of central nervous system properties, including specific neurotransmitter systems.²³ It is conceivable that alterations in descending (e.g., olivocochlear, acoustic reflex) pathways that might normally protect the ear are caused by genes. For example, inbred strains differ with respect to transmitter systems such as acetylcholine,²⁴ which may be involved in efferent auditory pathways.

Other centrally acting gene effects might cause the central auditory system to become more or less vulnerable to denervation secondary to noise-induced cochlear damage. For example, Willott et al.²⁵ observed less severe changes in the cochlear nucleus of CBA mice with severe noise-induced cochlear damage than in C57 mice with less severe, genetically determined cochlear pathology. In other words, genetic influences on the severity of NIHL need not be limited to the ear, but could extend to the central auditory system.

Genes might even alter psychological factors that could increase the risk of noise damage. For example, factors that influence an individual's tolerance of noisy conditions (e.g., auditory discomfort levels, physiological stress mechanisms, intelligence) could be strongly influenced by genes. Such effects could determine the degree to which intense noise is avoided.

It is clear that many potential routes exist by which genes might modulate susceptibility to

NIHL, and that little is known about any of these. The remainder of this chapter focuses on one approach that is beginning to provide new information on genes and susceptibility to noise—the use of inbred, F1 hybrid and backcross strains of mice exposed to intense noise.

NIHL in Inbred Strains of Mice

Peripheral function develops in an apparently normal fashion in young adult C57 mice (i.e., 1–2 months of age) although small elevations of thresholds for high frequencies may be first observed by about 2 months of age.^{26–30} By 4–6 months of age, thresholds for high frequencies (e.g., > 20 kHz) become elevated significantly and by 1 year of age, high-frequency losses are severe, and middle to low frequencies are also affected.^{30–39} Cochlear histopathology appears to be responsible for the hearing loss and has been well documented.^{31–33,38,40,41} Prior to 2–3 months of age little cochlear pathology is evident in C57 mice, although mild degeneration of outer hair cells (OHCs) has been observed in the extreme base of the cochlea at this age.^{30,38} By 6 months, degenerative changes of the organ of Corti (e.g., distortion, clumping, and loss of OHCs) are quite evident and are most pronounced in the basal turn.

Recent evidence indicates that C57 mice are also more susceptible to NIHL than the CBA strains that hear normally to an advanced age.^{26,33–36} Shone et al.³⁷ showed that C57 mice at 6 months of age, when high-frequency threshold elevations and loss of OHCs are significant, were more susceptible to NIHL than comparably aged CBA mice.

Li and colleagues^{26,38,42–44} also studied the effects of traumatic noise exposure in the CBA and C57 strains of mice. Anesthetized mice were exposed to traumatic noise (2–7 kHz, 120 dB sound pressure level, SPL, for 5 minutes) at 1, 2, 3, and 5 months of age. Both strains of mice were most susceptible to NIHL at 1 and 2 months of age, exhibiting permanent threshold shift (PTS) of 20–30 dB 1 month after exposure. After noise exposures at 3 and 5 months of age, the CBA mice exhibited very

little PTS whereas the C57 mice persisted to have PTS of about 20 dB. The assessments of noise-induced PTS were made for stimuli of the midhearing range for mice (8–12.5 kHz, $\frac{1}{3}$ octave bands). There was considerable variability within and between the strains of mice. These studies demonstrate that the 1–2-month-old CBA mice are susceptible to this traumatic noise exposure; their ears mature by 3 months of age so that they are much less susceptible. The C57BL/6J mice remain susceptible to the traumatic noise and/or they exhibit less recovery from temporary threshold shift (TTS) than the CBA mice between 24 hours and 1 month after noise exposure. These studies afford limited genetic interpretation for two reasons: brief traumatic exposures may produce greater and more variable injury; the spectrum (2–7 kHz) of noise exposure was below the measured TTS and PTS. Our C57BL/6J mice exposed to 110 dB for 1 hour showed no PTS for a lower (< 10 kHz) spectrum noise, but they did exhibit PTS at 16 kHz for a 10–20 kHz noise exposure.

The series of experiments to be described extends the findings with C57 and CBA mice. An attempt was made to use relatively moderate noise exposures within a range of intensities and durations sufficient to produce a differential effect between two or more strains or genotypes of mice. Furthermore, we used 3–4-month-old C57 mice (when age-related cochlear pathology is still minimal) so that the effects of noise exposure would not be confounded by the effects of AHL.

Background: Genetics of Age-Related Hearing in Mice

The study of any genetic effects requires more than the comparison of two inbred strains of mice with different phenotypes, including AHL or NIHL. Erway et al.⁴ were able to follow hearing loss in a genetic study of aging that involved 5 inbred strains of mice plus the 10 F1 hybrid strains derived from them. Fortunately the inbred strains of mice included a normal hearing CBA/H-T6J strain, the C57BL/6J strain, and three other inbred

strains (DBA/2J, BALB/cByJ, and WB/ReJ) that also exhibited AHL. The 10 F1 hybrid strains of mice exhibited either normal hearing to 23 months of age or patterns of AHL. From these patterns of normal hearing and AHL we deduced the probable existence of three different recessive genes for AHL, one of which is carried by the C57 strain of mice.

In an independent study L.C. Erway, K.R. Johnson, S.A. Cook, P. Ward-Bailey, and M.T. Davisson (unpublished data, 1995) demonstrated that a single gene from C57 mice segregates among backcross progeny as a recessive gene and maps to Chromosome 10. It was postulated that the AHL, and perhaps also the increased susceptibility to NIHL in these mice, may be due to homozygosity for the gene designated *Ahl/Ahl* on Chromosome 10.

NIHL in Inbred and Hybrid Strains of Mice

The effects of NIHL were investigated in the inbred CBA/CaJ and C57BL/6J strains and in two F1 hybrid strains of mice that, respectively, possess the heterozygous (+/*Ahl*, CBAx57F1) and homozygous (*Ahl/Ahl*, C57xDBAF1) genotypes. All inbred and all F1 hybrid strains of mice are genetically homogeneous within each strain, all mice of each strain being as genetically alike as identical twins. Therefore, if there are any clearcut differences for NIHL between the two inbred strains and between the two hybrid strains of mice, this would support the hypothesis that the differences in NIHL are attributable to a major gene, the putative *AhL/AhL* genotype.

L.C. Erway, Y.W. Shiau, and R.R. Davis (unpublished data, 1995) exposed each of these four strains of mice to a broadband noise (5–31 kHz) at 110 dB SPL (re: 20 μ Pa) for 1 or 2 hours and compared them with unexposed controls. Twenty-four mice of each strain were screened for normal hearing before exposure and tested by ABR thresholds (clicks, 8, 16, and 32 kHz tone pips) after exposure (2–7 hours, 1 and 3 days, 1 and 2 weeks, 2 and 3 months).

All mice of the C57BL/6J inbred and of the C57xDBAF1 hybrid strains exhibited significant PTS after noise exposure. Furthermore, these two strains of mice did not differ significantly from each other. The NIHL in these two strains of mice was greatest at 16 kHz, less at 32 kHz, and least at 8 kHz and for clicks. Whereas the CBA/CaJ inbred and CBAxC57F1 hybrid strains of mice exhibited TTS, they recovered completely by 3 or 7 days postexposure. It was concluded that mice of the C57BL/6J inbred and C57xDBAF1 hybrid strains (both *Ahl/Ahl*) were much more susceptible to NIHL than mice of the CBA/CaJ (+/+) and CBAxC57F1 (+/*Ahl*) strains. Moreover, the two *Ahl/Ahl* strains of mice exhibited AHL between 5 and 8 months of age in addition to the already existent NIHL. The time course of AHL was consistent with that observed in the unexposed control mice. All of these results suggest that susceptibility to NIHL and to AHL have a common genetic basis.

Segregation for Gene Causing NIHL in Backcross Mice

The results described above support but do not prove that the observed differences in NIHL may be due to the *Ahl/Ahl* genotype associated with Chromosome 10. Such genetic evidence and mapping of genes requires appropriate genetic crosses. We chose the simplest and most relevant cross, namely backcross of the CBAxC57F1 hybrid mice (+/*Ahl*) to the C57BL/6J inbred mice (*Ahl/Ahl*). Such backcrosses are expected to yield $\frac{1}{2}$ +/*Ahl* and $\frac{1}{2}$ *Ahl/Ahl* progeny. Whatever differences in NIHL may be attributable to a major gene, they should be detectable as different phenotypes (presence or absence of PTS) associated with these two genotypes. However, other genes (hundreds of which differ between these two strains) will also segregate and assort, producing extensive heterogeneity for the genetic background of these progeny.

Unexpectedly, none of these backcross progeny exhibited significant PTS after 1 hour

of exposure to 110 dB. We had to reexpose these mice to 110 dB to obtain PTS in approximately half of the backcross progeny. There was, however, a clearcut disparity in the rate of recovery from TTS among subgroups of the backcross progeny.

Noise exposure of 110 dB for 8 continuous hours yielded two subgroups of mice differing with regard to the rate of recovery from TTS. (These differences were objectively determined by cluster analyses as demonstrated below and in Figure 6-1.) Approximately half of the backcross progeny exhibited a slower rate of recovery (days 1, 3, and 7 after exposure), with a persistent PTS (20–40 dB at 16 kHz). However, some of the mice with delayed recovery from TTS did recover between 7 and 14 days, thus exhibiting no PTS. Reexposure of these mice to 110 dB for 4 hours produced PTS to 14 or more days.

Mapping of Gene for Susceptibility to NIHL

A total of 91 of the backcross mice were exposed to noise and tested for NIHL. These mice were typed by X.B. Ling and G.A. Cortopassi (unpublished data, 1995) for two genetic markers on Chromosome 10. Sixty-seven (73%) of these progeny exhibited the expected parental genotypes and phenotypes:

1. Approximately half of these mice exhibited normal hearing (no PTS), and they were heterozygous for both markers of the CBA and C57 strains; all of these normal hearing mice were thus concordant for heterozygosity for the two markers and for the AHL locus (+/*Ahl*).
2. The other half of these mice exhibited patterns of TTS and PTS (see cluster analysis) and were homozygous for both markers of the C57BL/6J strain; these mice were thus concordant for homozygosity for the two markers and for the AHL locus (*Ahl/Ahl*).

An additional 16 (17%) of the backcross progeny exhibited recombination (crossing

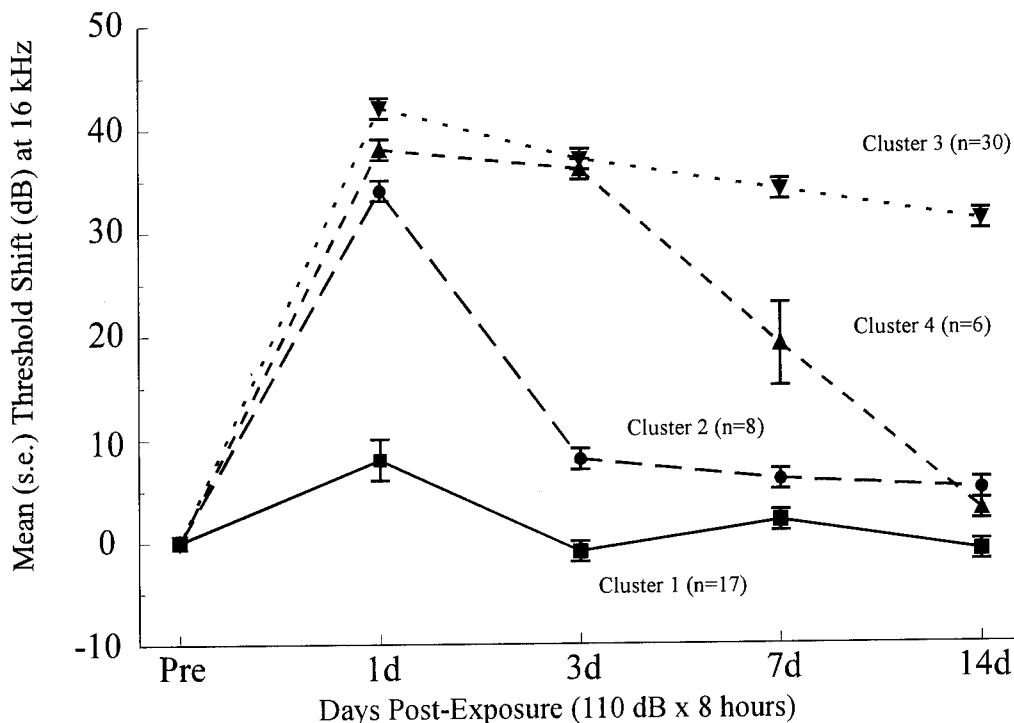


Figure 6-1 Cluster analysis for NIHL among backcross progeny (CBAxC57) F1xC57BL/6J. The threshold shifts (TSs) for 16 kHz were computed at 1, 3, 7, and 14 days after exposure to 110 dB for 8 hours. Four different patterns of NIHL were identified by cluster analysis. The means and SEs of each of the patterns are shown with the number of mice in each pattern. Mice in cluster 1 had the least TS at 1 day; mice in cluster 2 recovered by day 3. Mice in cluster 3 exhibited permanent TS; six mice in cluster 4 had similarly high TSs to day 3 but recovered to preexposure levels by day 14. Clusters 1 and 2 were collapsible providing candidates for the *+Ahl* genotype. Clusters 3 and 4 were collapsible providing candidates for the *Ahl/Ahl* genotype. Clusters 1 and 2 were not collapsible with clusters 3 and 4. The 25:36 ratio of *+Ahl* to *Ahl/Ahl* is not significantly different from the expected 1:1 (see text for discussion).

over) between the two molecular markers on Chromosome 10:

1. mice that exhibited normal hearing were concordant with heterozygosity for one of the two markers and for presumed heterozygosity for *+Ahl*;
2. mice that exhibited NIHL were concordant with homozygosity for one of the markers and for the presumed homozygosity for *Ahl/Ahl*.

The 90% of the backcross progeny described above establishes that a genetic factor contributes to susceptibility to NIHL and that such a factor is linked to two molecular markers on Chromosome 10.

Interactions Between Genotypes and With Environmental Noise

The relationships between genotype and phenotype are typically far removed from each other, especially for a phenotype as complex as normal hearing. The remaining 10% of the backcross progeny did not conform to the simplest hypothesis: seven mice exhibited significant PTS that was discordant with the observed heterozygosity for both markers and thus being putatively heterozygous for *+Ahl*. Two additional mice were homozygous for both markers, but they did not exhibit any PTS, being thus discordant with their putative homozygosity for *Ahl/Ahl*.

These nine mice might have been considered to be the result of double crossovers occurring simultaneously between the three loci; however, such probability is less than 0.01 and could not have occurred more than once or twice for the nine mice that were discordant between the two markers and the hearing phenotype.

The two *Ahl/Ahl* mice without PTS are most simply explained by the fact that some other *Ahl/Ahl* mice required reexposure to noise to exhibit PTS. It appears that 110 dB for 8 hours is not sufficient to produce PTS in all of the *Ahl/Ahl* backcross progeny.

The seven mice putatively *+ / Ahl* but exhibiting PTS are more difficult to explain; each of these exhibited PTS from the single, 8 hour exposure to 110 dB. None of the 16 CBAx57F1 hybrid mice (*+ / Ahl*, exposed to 110 dB for 1 or 2 hours in the previous study) exhibited any PTS. None of the F1 hybrid mice were subjected to greater exposures. However, the backcross progeny required 8 hours or more exposure to obtain significant PTS in all of the *Ahl/Ahl* progeny.

It is possible that 110 dB for 8 hours exceeded the threshold at which these *+ / Ahl* backcross mice are susceptible to NIHL. We examined this possibility by subjecting *+ / Ahl* mice to lower levels of noise exposure. As noted above, some of the *Ahl/Ahl* mice failed to exhibit PTS after exposure to 110 dB for 8 hours, requiring further exposures to obtain PTS. Therefore, there appears to be some overlap in the range of sensitivity of the two genotypes to levels of noise exposure. There was no such overlap among the two inbred and two F1 hybrid strains; all 16 individuals within each resistant strain (*+/+* or *+ / Ahl*) exhibited no PTS, and all 16 mice within each susceptible strain (*Ahl/Ahl*) exhibited PTS.

A possible explanation for the variability of NIHL among the *+ / Ahl* and *Ahl/Ahl* progeny is the heterogeneous genetic background among all backcross progeny. By contrast, the F1 hybrid strains of mice have a highly homogeneous genetic background. If any other gene(s) within the CBA or C57 strains of mice may further modify the susceptibility to NIHL, such gene(s) would be segregating and

assorting also with the *Ahl* alleles. Such other genotypes could either ameliorate the effect of *Ahl/Ahl* or exacerbate the effect of *+ / Ahl*. Such modifier genes have been demonstrated often in other genetic organisms and systems.

In conclusion, the initial studies of susceptibility to NIHL indicate that the inbred and F1 hybrid strains clearly differ from each other, consistent with different genotypes (*+ / Ahl* vs. *Ahl/Ahl*). It remains to be determined if other genes affecting AHL may cause mice to be differentially susceptible to NIHL. Initial genetic mapping studies among backcross progeny indicate that the AHL locus on Chromosome 10 has a major effect on susceptibility to NIHL, but other background genotypes may modify the effect on NIHL.

Cluster Analysis for Identifying the NIHL Phenotype

The backcross mice are genetically heterogeneous in contrast to the genetically homogeneous inbred and F1 hybrid strains of mice. The only way to classify particular genotypes among backcross progeny is to identify valid criteria for observed phenotypes as noted above. Sixty-one backcross progeny of comparable ages (4–5 months) were exposed to 110 dB noise for 8 hours and tested for ABR thresholds at 1, 3, 7, and 14 days after exposure. Based on the thresholds shifts (TS) observed between the pre- and postexposure thresholds, all 61 mice were clustered (Ward's minimum variance) into four groups. These clusters are shown in Figure 6-1 with TS means \pm SE for the 16 kHz test stimulus. Clusters 1 ($n = 17$) and 2 ($n = 8$) exhibited the least TS but were permissibly combined, thus representing the *+ / Ahl* genotype. Clusters 3 ($n = 30$) and 4 ($n = 6$) were separated on the basis of the recovery between days 3 and 14 of the mice in cluster 4; mice in clusters 3 and 4 were permissibly combined, thus representing the *Ahl/Ahl* genotype.

The genetic mapping data indicated that 90% of these backcross mice were concordant between the two observed NIHL phenotypes (clusters 1 and 2 vs. 3 and 4) and the probable genotypes, *+ / Ahl* and *Ahl/Ahl*. However, the

genetic mapping data also indicated that seven mice from cluster 3 were probably $+/Ahl$ and thus discordant with the NIHL phenotype; contrariwise two mice from clusters 1 and 2 were probably Ahl/Ahl and thus discordant for the lack of NIHL. The 90% concordance establishes a major effect of the genotype on Chromosome 10; the 10% discordance allows for either stochastic effects on NIHL or for other genetic modifiers of NIHL within the $+/Ahl$ and Ahl/Ahl genotypes. The genetic heterogeneity among backcross progeny versus homogeneity for inbred and hybrid mice favors the effect of other genetic modifiers on NIHL.

Relevance to NIHL in Other Species

Humans exposed to a lifetime of noise in the workplace have shown markedly different responses from virtually no hearing loss to extensive hearing loss. Chinchillas exposed to comparable levels of noise have frequently exhibited large differences in NIHL and hair-cell loss.⁴⁵⁻⁴⁷ Chinchillas show varying patterns of recovery from impact noise⁴⁸ that resemble recoveries of our backcross mice (see cluster analysis). Guinea pigs exposed to well-controlled levels of noise have shown large interanimal variation in NIHL and damage to the cochlea.^{49,50}

Humans, chinchillas, and guinea pigs are not highly inbred. Despite small founding populations or breeding colonies, few mammalian species other than mice have been highly inbred (by pedigree to hundreds of generations). Random-bred or outbred populations typically retain a large amount of genetic heterozygosity; they have many loci that are heterozygous for two or more alleles. Heterozygosity for any loci that may affect physiological processes could alter the susceptibility to NIHL. It is certainly possible that large variations observed among individuals of the same species, breed, or strain, may be due to certain genotypes at one or a few genetic loci. The possible genetic basis for such variation in chinchillas, guinea pigs, or other animal models, can only be ascer-

tained by appropriate screening and breeding programs.

The possibilities for genetic variability in humans for susceptibility to NIHL are considerable, but humans are less amenable to study than are animal models. Given the relationship shown to date between susceptibility to NIHL and genetic predisposition to AHL, or presbycusis, consideration should be given to investigating NIHL in human families with predisposition to presbycusis.

When any of the genes affecting AHL and/or NIHL in mice can be identified and mapped to precise regions of the chromosome, they may provide models for understanding both phenomena in humans. The homologies between human and mouse chromosomes are largely known. Any mouse gene may be useful in identifying and mapping homologous human genes. Any genes identified in the mouse may lead to understanding of the mechanisms underlying normal homeostatic mechanisms of the inner ear.

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Chapter 7

Effects of Acoustic Overstimulation on Distortion-Product and Transient-Evoked Otoacoustic Emissions

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Otoacoustic emissions (OAE, discovered by Kemp¹) are believed to originate from electro-mechanical processes in the outer hair cells (OHC) of the organ of Corti. These processes involve OHC motility²⁻⁴ and are supposed to be a part of the so-called cochlear amplifier.⁵ It is widely held that they contribute to the high sensitivity and frequency selectivity of a normal cochlea. When OHCs are impaired, the sharp tuning normally found in auditory nerve fibers coming from the damaged regions disappears and a 40–60 dB loss of sensitivity can be observed. The presence or absence of evoked OAEs (EOAEs) exhibits parallel behavior. Transient-evoked and distortion-product OAE (i.e., TEOAEs and DPOAEs) can be recorded from almost any normal cochlea, provided middle ear transmission is normal. In contrast, both types of signals tend to disappear whenever a hearing loss due to OHC dysfunction exceeding 30–40 dB is observed in the frequency range where they would be normally expected. Neonatal hearing screening is based upon this finding, and has proved to be reliable enough that recently TEOAEs have been recommended as a systematic screening test.⁶

Acoustic overstimulation is one of the main causes of acquired OHC damage.^{7,8} Therefore, EOAE has been proposed as a fast, sensitive, objective tool for assessment of the effects of noise exposure. Indeed, several lines of reasoning suggest that EOAE measurements might provide an interesting alterna-

tive to conventional pure-tone audiometry, especially for monitoring early cochlear alterations in subjects exposed to intense sounds. For several reasons, it would be important to design an objective method that would be more effective than pure-tone audiometry. First, normal hearing thresholds as well as other psychoacoustic variables obtained in a clinical setting can be observed in animal experiments in association with large sensory cell losses.⁹ However, such tests are time consuming and require the cooperation of subjects; thus they cannot be routinely performed in human studies.¹⁰ Second, objective methods of assessment may prove to be very important for medicolegal purposes, especially for quantitative evaluation of impairment resulting from noise-induced hearing loss (NIHL). Third, EOAEs most probably arise as a result of a highly tuned mechanism. Recall that OAEs were discovered by Kemp¹ during an investigation of the fine structure of auditory thresholds and that the association between OAEs and threshold microstructure has been confirmed.¹¹ It is tempting to speculate that any damage to EOAE sources will be reflected in EOAEs in a frequency-specific manner. This should mean that not only the frequency spectra of the EOAEs should correlate with the spectra of the eliciting sounds, but also that the EOAEs should reflect the mechanical properties of OHCs tuned to these frequencies. Although this last issue has been addressed by recent studies, there has been

no direct proof. However, EOAEs do appear to be an effective frequency-specific tool for detecting hearing losses larger than about 40 dB, with a sensitivity and specificity similar to other routine audiometric methods (where specificity is the ratio of numbers of actual impaired ears to ears without EOAE). However, several attempts^{10,12,13} to use EOAEs as an early predictor of moderate cochlear damage have failed, regardless of the cause of the OHC alterations. These studies raised serious concerns as to the possibility of the early detection of cochlear damage because threshold elevations were clearly present while EOAEs remained in the normal range. In addition, Lutman and Saunders¹⁴ could not find any peculiarities in the TEOAEs in a large series of patients having normal pure-tone thresholds that complained of "obscure auditory dysfunction," even though their symptoms most likely were the result of some OHC dysfunction. Other experiments, however, did suggest that EOAE changes might anticipate pure-tone audiometric changes in the case of aging^{15,16} or NIHL.^{17,18} However, in conditioning exposures using moderate levels of noise^{19,20} or in the assessment of individual susceptibility,^{21,22} promising results using the DPOAEs have been described.

The number of different parameters that need to be specified when defining an EOAE paradigm is very large, as is the number of measurable characteristics of the resulting signals. It is not possible to perform a thorough evaluation of the effects of all possible combinations of parameters in a single experiment. However, the importance of parameter choice, for example the number of test frequencies, ratio of primary frequencies, and levels in a DPOAE experiment, was described by Probst et al.²³ The choice of EOAE parameters may explain some of the discrepancies between the previously mentioned studies. Another important challenge is to improve our knowledge about the basic features of EOAE physiopathology in animal models. Clinical research has provided a wealth of data on normal and impaired human ears, but normative data bases exhibit a large variance, and

EOAE recordings pre- and posttrauma are seldom available in human subjects.

The aim of this chapter is to present a discussion of currently available data for use in evaluating the advantages and limits of EOAEs in cochlear assessment. Several paradigms are proposed for improving EOAE detection and sensitivity to minute OHC damage. With the availability of several commercial devices^{24–26} that enable fast, reliable, and routine recording of TEOAEs and DPOAEs, the field has become very active in the last few years. As a consequence, this will not be an exhaustive review.

Basic Techniques, Interest, and Limits

The basic techniques used for eliciting and detecting EOAEs have been extensively described.^{25,27} Therefore, they will not be described in detail in this chapter. To summarize, two categories of EOAEs are detectable in almost all normal mammalian ears, namely TEOAEs²⁸ and DPOAEs.^{29,30} Both can be used for evaluating the status of the cochlea. Transient EOAEs can be elicited either by short clicks (typical duration: 80 microseconds for the electrical signal sent to the earphone) or tone bursts with a narrower frequency spectrum. The duration of the EOAE response is longer in humans and primates than in animals such as rodents. Typical responses in a normal ear are composed of a unique and highly reproducible combination of frequency components in the 0.5–6 kHz range. Several experiments^{31,32} have confirmed that TEOAE responses to broadband clicks are a superposition of responses to tone bursts, equating for the respective levels and spectra of the stimulus. For this reason, the results obtained with any kind of TEOAEs can be discussed together. The main technical limitation of TEOAE techniques arises from the short latency of the high-frequency components. Above 4–6 kHz, the latency is shorter than 2.5 milliseconds, which makes it impossible to separate stimulus ringing from true EOAEs. Thus the direct exploration of a large range of

frequencies that would be required assessing cases of NIHL does not seem possible.

In contrast, DPOAEs are produced at $2f_1-f_2$, $2f_2-f_1$, f_2-f_1 , etc., when two primary tones at f_1 and f_2 (with $f_2 > f_1$) are presented simultaneously in the external ear canal. The cubic distortion tone $2f_1-f_2$, being most prominent, has been extensively studied. According to many studies, $2f_1-f_2$ is generated in the region of the primary tones,³³ hence it should be possible to monitor the status of the cochlea around this location. f_1 and f_2 can be almost arbitrarily chosen³⁴ up to 100 kHz provided the difficulties associated with sound calibration in the outer ear canal are addressed.^{35,36} Furthermore, the stability of the $2f_1-f_2$ DPOAE in test-retest experiments has been demonstrated,³⁷ while the quadratic combination tone at f_2-f_1 , which can be large under certain conditions, has been shown to present a poor short-term stability.³⁸ A very important point has been made about the maximum primary level giving rise to truly "active" DPOAEs^{39,40}; when the level of primary tones exceeds 70 dB sound pressure level (SPL), cubic distortion tones can be detected even in the absence of OHC. The idea that there are two discrete sources of DPOAEs, one of them being less dependent on OHC activity, has been suggested.⁴¹

Basically, all types of EOAEs have the same origin in that they arise as a by-product of the electromechanical transduction stage in healthy OHCs. However, the details of their generation in the organ of Corti and their backward propagation along the basilar membrane remain almost totally unknown. Long ago, after anticipating the existence of active mechanisms in order to explain the normal cochlear tuning, and predicting that spontaneous otoacoustic emissions might result from such mechanisms, Gold also predicted that TEOAE cannot exist in a perfectly orderly system.⁴² Following this idea, Shera and Zweig⁴³ and Wit et al.⁴⁴ proposed that a certain amount of disorder, either anatomical or functional, should be assumed to exist in the cochlea for TEOAEs to be detected with their characteristic temporal and spectral patterns.

Incidentally, TEOAEs also exhibit a large degree of saturating nonlinearity as a function of stimulus level. Nonlinear growth is used by detection systems as a criterion for separating true TEOAEs from acoustic artifacts. Nonlinearity and disorder in cochlear micro-mechanics also influence DPOAE properties. Some particularities of DPOAEs seem to reflect the presence of some disorder in their generating system,⁴⁵ but this phenomenon does not seem as essential as in the case of TEOAEs. Obviously, nonlinearities are indispensable for DPOAE generation but their intimate nature is as yet unknown. Kirk and Johnstone⁴⁶ proposed that quadratic and cubic distortion tones, respectively at f_2-f_1 and $2f_1-f_2$ may originate from distinct physiological mechanisms. Although such issues remain highly speculative, it is important to keep them in mind because they may help to understand some possible differences in sensitivity to cochlear dysfunction among the different types of EOAEs.

In summary, in almost all studies of relationships between EOAEs and cochlear pathology, the working hypothesis is the following: correlations are supposed to exist between the characteristics either of TEOAEs at some frequency f , or DPOAEs produced by primary tones around f , and the status of the cochlea at the place tuned to the frequency f . The choice of the type of EOAE suitable for an experiment is guided by the different technical limitations of these signals, that is, amplitude, signal-to-noise ratio, and more generally, artefact elimination, reproducibility, and frequency range. Both types of EOAEs have a similar drawback when utilized in physiopathological studies when no reference recording "before lesion" is available. The large variance found in samples of normal ears hampers any early detection of abnormal EOAE change. Animal experiments aim to overcome these difficulties.

TEOAEs in NIHL

Few data are available to date on human studies of NIHL, partly because beginning

NIHL is found around 4–6 kHz in humans and TEOAEs can be absent in this frequency range even in normal ears, especially in adults. The situation is even worse in animal studies, owing to the very short latency of these emissions that makes them difficult to separate from stimulus artefact.

TEOAE Characteristics of Interest

TEOAE responses are said to be idiosyncrasic, that is, their time and frequency patterns are unique in a given ear and stable in the absence of cochlear or middle ear insult. Their amplitude, signal-to-noise ratio, growth function, and detection threshold at every frequency can be measured and summarized in the TEOAE frequency spectrum for every stimulus level. In addition, reproducibility, defined as the cross correlation between repeated measurements, can be useful.²³

Smurzynski and Kim⁴⁷ analyzed the conditions for building a human data base of normative TEOAE spectra (in 48.8 Hz frequency bands). Responses corresponding to 10th and 90th percentiles of their sample of normal ears were found in the range (+1, –21 dB) at 1 kHz, or (–10, < –30 dB) at 4 kHz for example (see their figure 2). For the 10th percentile, their frequency range was 0.5–6 kHz, but was restricted to 0.8–3.7 kHz for the 90th percentile. These normal variations are conspicuously large. Moreover, it is well known that several hundred hertz wide notches can be found in TEOAE spectra of normal ears. Therefore, it is clear that comparisons between the responses of a human subject following a noise exposure and such a data base may not be very sensitive in the detection of early noise-induced damage. Moreover, the question as to how to decide that a subject can be included in a normal data base is difficult because it is seldom possible to determine whether or not the subject has been exposed to harmful sounds, especially considering that pure-tone thresholds may not be a sensitive enough predictor of early damage to OHCs.

Another approach has been proposed by Prieve et al.⁴⁸ who applied the theory of statistical decision to evaluate what values or com-

binations of TEOAE parameters result in the best prediction of hearing threshold shifts. They concluded that TEOAE level, TEOAE-to-noise ratio, and percent reproducibility were equally effective. Frequency-specific frequency bands could be identified, that is, alterations of TEOAE properties in a frequency band are correlated with audiometric changes in the same band.

Applications to Humans With NIHL

A common observation in clinical studies is that the shapes of TEOAE frequency spectra roughly correspond to the frequency range with a near normal pure-tone audiogram.⁴⁹ This means that the highest frequency found in the residual TEOAE components and the lowest frequency with a hearing threshold better than 20 dB hearing level (HL) are often close to each other. It must be kept in mind that the usual shape of audiograms with NIHL is either a ski-slope, that is, hearing loss increasing with frequency, or a notch centered around 4 kHz. Thus, multiple-correlation studies between the TEOAE spectrum and audiogram may seem to reveal statistically significant frequency-specific correlations. However, such correlations are not valid because the audiometric variables are not independent, thus no conclusion can be drawn as to the frequency specificity of TEOAE recordings. In other words, such correlations do not prove that TEOAEs are good predictors of the progression of OHC damage due to noise; on the contrary, more valid statistical methods suggest that TEOAE properties at frequency f are best correlated to audiometric thresholds at frequencies higher than f .^{17,18}

More powerful analyses can be performed when subjects exposed to noise can be tested at the beginning and end of an exposure session. In such situations it has been shown that TEOAE testing might be more sensitive than pure tone audiometry.⁵⁰ Following this idea, LePage and Murray¹⁵ carried out a statistical analysis of TEOAE in more than 1000 subjects, and deliberately restricted it to the global TEOAE amplitude (coherent output power,

defined as the product of echo level and reproducibility coefficient). They showed a clear decrease of average echo levels with age, in the absence of auditory threshold changes. Because age is presumably associated with a higher probability of OHC damage due to acoustic overstimulation, their results could be considered an indirect demonstration of a sort of anticipation of cochlear damage by TEOAEs.

An alternative interpretation of these data can be proposed if one admits that some characteristics of TEOAEs may depend on the cochlear status at remote basal places. In NIHL, these places likely exhibit some degree of damage earlier than the cochlear places tuned to the frequencies of emissions. Several lines of evidence do suggest that the whole cochlear base may influence the generation and propagation of TEOAEs.^{51,52} We designed an experiment in the guinea pig in an effort to evaluate the influence of basal cochlear damage on TEOAE amplitudes at lower frequencies.

Application to Animals Exposed to Loud Sounds

TEOAEs could be detected in normal guinea pigs but an effective artefact rejection had to be implemented because the duration of the emission responses is much shorter in rodents than in primates. Special care was also taken to ensure a tight fit of the EOA probe tip in the outer ear canal. Guinea pigs were anesthetized and a round-window electrode was installed for monitoring compound action potentials (CAPs) during the experiments. CAPs were evoked from 1 to 32 kHz using calibrated tone pips with varying levels. Visual detection thresholds were estimated. TEOAEs were elicited by 80 microsecond clicks at 65 dB peak-equivalent SPL. TEOAE spectral components were clearly visible from 1.5 to 5 kHz in all of the 18 tested animals. The preexposure TEOAEs and CAP thresholds served as a reference. The guinea pigs were exposed to loud pure tones for 5–10 minutes at 95–105 dB SPL. The frequency of overstimulation was chosen between 3 and 10 kHz. Subsequent temporary threshold shifts (TTS) were found in the ex-

pected frequency range, i.e., with a maximum at about half an octave above the frequency of exposure. This maximum threshold shift ranged from 20 to 50 dB. Partial or total recovery could be obtained after a few hours, but was slow enough to allow CAP and TEOAE measurements to be made under relatively stable conditions. It is important to note that in most cases, no TTS was found at the frequencies of emissions. However, significant decreases in the amplitudes of TEOAE frequency components were observed. The amount of change could be as large as –17 dB, with an average value of –6 dB and an SD of 4 dB. Interestingly, increases in TEOAE occurred during CAP threshold recovery, and TEOAEs returned to preexposure levels whenever a complete recovery was observed.

Figure 7-1 shows a typical example of the observed changes. The TEOAE frequency spectrum of this guinea pig was measured before any exposure to loud sound and is represented as a dashed line between 1.5 and 5 kHz. The background noise level was always less than –25 dB SPL. The CAP thresholds were normal between 1 and 32 kHz and the corresponding acoustic levels were arbitrarily set at 0 on the reference CAP audiogram. Then, the animal was exposed to a pure tone at 8 kHz (105 dB SPL, 5 minutes), using the same loudspeaker and probe as for TEOAE and CAP measurements. Therefore, the probe was not removed from the ear canal. Immediately after exposure, the animal's CAP thresholds and TEOAE were measured twice and proved to be stable. CAP thresholds remained stable during 30 minutes before some recovery took place. Postexposure TEOAE and CAP recordings are represented in Figure 7-1. The postexposure frequency spectrum of TEOAE reveals a decrease of 2–3 dB in all frequency components relative to the preexposure control. Nonetheless, no CAP threshold change was found at the frequencies of TEOAEs (short arrows in Figure 7-1).

Because it was impossible to find any frequency-specific correlation between functional changes and TEOAE properties and the exposure did not induce threshold changes at the frequencies of emissions, we looked

for possible relationships between TEOAE changes and the extent of CAP changes at the high frequencies. The width of the audiometric notch, defined as the width in octaves of the frequency interval with a TTS larger

than 10 dB, was evaluated. The percentage of basal cochlear length with an unaltered function, the residual base (RB), was deduced from this width for every frequency component of the TEOAE. A highly significant linear regression was found between ΔEOAE , defined as the amplitude change of a TEOAE component, and RB.

$$\Delta\text{EOAE (dB)} = 0.70 \times 20 \log \text{RB} - 0.63$$

$$(r = 0.69, n = 52 \text{ cases}).$$

This relationship is quite surprising because it shows that very remote parts of the basal cochlea may play an important role in the determination of TEOAE amplitudes. The oversimplified assumption that every part of the basal cochlea may contribute to an emission with the same weight would lead to the relationship $\Delta\text{EOAE} = 20 \log \text{RB}$. The linear regression presented above is not very different. The evaluation of RB is significant because both the total length of active basal cochlea and the actual length of the cochlea along which damaged OHC are distributed, cannot be accurately measured in the absence of histological data. Nevertheless, it is easy to check that the RB is highly robust whatever the precise definition of the notch. It is also noteworthy that the influence of RB is fairly weak inasmuch as a change of 50%, produced for example by an audiometric notch of about 2 octaves width, would only result in a decrease in TEOAE amplitude of about 4.2 dB. Note that we obtained similar results using narrowband stimuli for eliciting EOAEs.⁵³

An alternative explanation to the apparent sensitivity of TEOAEs to cochlear basal damage would be that ΔEOAE actually results from a minute amount of OHC dysfunction at the place tuned to TEOAE components. This OHC damage might be insufficient for CAP thresholds to be altered. However, it is possible that ΔEOAE would be strongly correlated with OHC alterations at more basal places, thereby explaining the previously described relationship. Morphological and DPOAE controls performed in a guinea pig experiment using similar types of overexposure did not confirm this tentative explanation.⁵⁴ Furthermore, it is unlikely that an over-

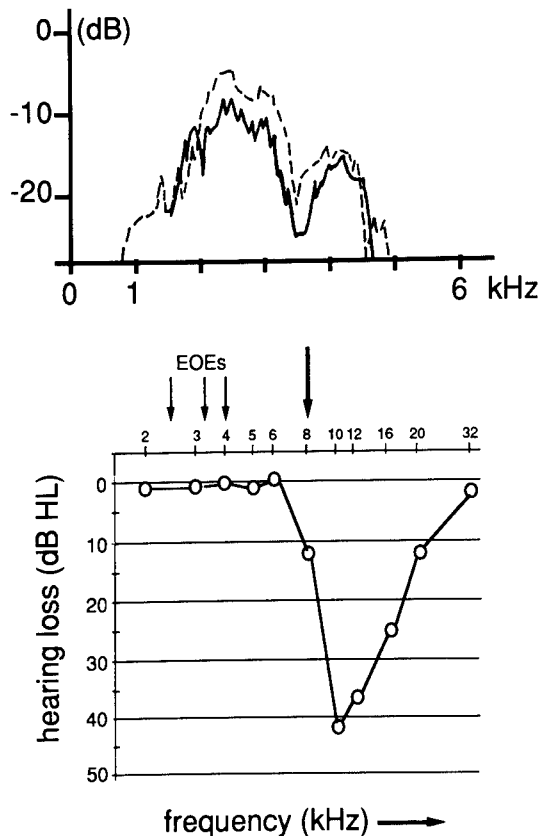


Figure 7-1 (Top) TEOAE spectrum in a guinea pig's ear—before (---) and after exposure to a loud tone at 8 kHz (105 dB SPL, 5 minutes). Background noise is not represented for sake of clarity but was equal to or lower than -25 dB at all frequencies. Between 1 and 1.5 kHz, no emission was found above the noise floor after exposure; thus the thick line representing their spectrum is interrupted. Three TEOAE frequency components could be identified around 4, 3.2, and 2.5 kHz. Their amplitude was decreased by 1–3 dB after exposure. **(Bottom)** At the same time, CAP threshold shifts after overexposure were determined between 2 and 32 kHz. The maximum shift was at 10–12 kHz as expected owing to the frequency of exposure (long arrow). No threshold shift was found at the frequencies of evoked emissions (shorter arrows). Nevertheless, TEOAEs were clearly modified.

exposure to an 8 kHz pure tone at 105 dB over a few minutes would damage the cochlear place tuned to 2 kHz. Anyway, even histological controls might fail to reveal minute functional alterations in OHC so that it is difficult to rule out the hypothesis that TEOAE changes are simply a reflection of local damage.

Whatever the interpretation, two conclusions can be drawn from such an experiment. First, the current knowledge of the detailed mechanisms of TEOAE generation is far from sufficient to provide a framework for understanding noise-induced TEOAE changes. We can only propose a phenomenological model based on statistical analysis. Second, such observations suggest that TEOAE can be useful for monitoring early high-frequency cochlear damage, but the expected effect is weak compared to intersubject variance. Thus it may be missed if the results of the exposed subject are just compared to a normal data base. Control preexposure recordings are necessary and should thus limit the possible clinical applications.

DPOAEs in NIHL

Several studies of noise injuries in humans as well as other mammals have been performed using DPOAEs.⁵⁵⁻⁵⁷ DPOAEs have several advantages. They have a large amplitude with a good signal-to-noise ratio in most laboratory animals. Although human DPOAEs have a smaller amplitude in general, they can be detected almost as easily as in rodents using some temporal averaging, and the frequency of primary tones can be chosen over a large range of frequencies including those that are most sensitive to a developing NIHL. A number of lines of reasoning indicate that the cochlear region tested by DPOAEs is fairly narrow in the region tuned to the frequencies of the primary tones.⁵⁸ Thus it should be possible to probe a large extent of the cochlea using DPOAEs.

The interest of DPOAE for assessing the cochlear status after noise damage seems clear when the resulting hearing loss is large enough (about 40 dB or more). A wealth of data collected in humans and animals in-

dicates the good frequency specificity of DPOAE for detecting NIHL and other causes of OHC dysfunction.⁵⁹⁻⁶⁴ However, several surprising results have been published showing that DPOAEs may be far less effective in cases of mild hearing loss^{12,13} or in cases of normal auditory thresholds in the presence of OHC damage.¹⁹ This failure to detect cochlear damage might be attributed to an insufficient analysis of the DPOAE data. The analysis is often restricted to plotting amplitude against frequency for a fixed level of primaries, or amplitude against primary level for a fixed primary frequency. Therefore, we shall attempt to describe some promising possibilities of more advanced DPOAE data analysis.

DPOAE Characteristics of Interest

Probst et al.²³ published an extensive review of DPOAE characteristics. Various combinations of primary tones can be used. For most species, the optimal ratio of primary frequencies f_2/f_1 seems to be between 1.20 and 1.30, that is, the range where the largest DPOAE amplitudes are obtained.³³ The level of primary tones must not be too large (typically less than 70 dB SPL), otherwise part of the DPOAE response may not represent OHC nonlinearities.³⁹⁻⁴¹ The difference between primary levels, namely $L_2 - L_1$, may vary between 0 and -25 dB.

Amplitude and signal-to-noise ratio are the most straightforward parameters to measure for given frequencies of the primary tones. The DPOAE growth function can be plotted against primary levels, thereby allowing a detection threshold and slope of the DPOAE input/output function to be estimated. A salient feature of cochlear DPOAEs is that this slope seldom exceeds 1 dB/dB increase of both primaries, whereas "passive" DPOAEs, as those produced in a dummy ear at high primary levels, exhibit a typical slope of 3 dB/dB, as expected for the cubic distortion product.

Two other important characteristics of DPOAEs are more difficult to obtain. As already mentioned, the source of DPOAE is considered to be around the place tuned to f_2 ; thus if the frequency of primary tone f_1 ($< f_2$) is

swept while f_2 is kept constant, then the amplitude and phase of the DPOAE at $2f_1-f_2$ can be plotted against $2f_1-f_2$. The DPOAE amplitude presents a maximum when $2f_1-f_2$ is about half an octave below f_2 , that is, this corresponds to the optimal f_2/f_1 ratio of about 1.20. Allen and Fahey⁶⁵ and Brown and Williams⁶⁶ suggested that this maximum arises from a sort of second filter in the micromechanics of the organ of Corti. This filter might be associated with the coupling between the *stereocilia* and the tectorial membrane, thus its characteristics should provide some insight into this feature of cochlear mechanics.

Furthermore, the so-called "group delay" of DPOAE is defined as

$$t = (1/2\pi)(d\phi/df)$$

with $d\phi$ representing the variation of DPOAE phase when its frequency varies by df .⁶⁷ This group delay presumably consists of a sum of terms corresponding to the forward propagation of primaries to the place tuned to f_2 generation of the DPOAE by nonlinear interactions between f_1 and f_2 in the active cochlear filter. Then a backward propagation of the DPOAE is initiated. The second step seems largely predominant thus the value of DPOAE group delay provides clues as to the source of the DPOAE. The narrower the cochlear filter, the larger $d\phi$ and t . This test is routinely used in the ILO92 system designed by Kemp and Otodynamics Ltd. It enables one to separate the true physiological DPOAEs from passive DPOAEs.

Normative data bases for DPOAEs present features quite similar to TEOAEs.⁴⁷ The amplitude range between the 10th and 90th percentiles is about 10–12 dB whatever the primary frequencies. Close correlations were found between TEOAE and DPOAE at the frequencies for which both were detectable, for normal as well as impaired subjects.⁴⁷

Applications to Humans With Mild NIHL

To evaluate the ability of DPOAEs to detect limited hearing losses in humans, a preliminary experiment was performed with patients

suffering from uni- or bilateral high-frequency hearing loss due to occupational noise exposure. Thirty subjects took part in the study that was performed during routine clinical testing. Subject ages ranged from 22 to 45 years. Otoscopy, tympanometry, auditory brain stem response (ABR), and acoustic reflex studies confirmed that hearing loss was purely of cochlear origin. A Békésy automatic sweep-frequency audiogram was done in addition to standard pure-tone audiometry (octave steps between 0.125 and 1 kHz then half-octave steps between 1 and 8 kHz). This provided a more accurate characterization of each subject's audiometric notch.

DPOAEs were elicited by equilevel primary tones of 50–65 dB SPL synthesized by an Ariel DSP16 board (Cubdis® system implemented by J.B. Allen at Bell Laboratories). The frequency ratio f_2/f_1 was chosen at 1.22. f_2 varied from 10 to 1 kHz in 1 kHz steps. DPOAE amplitudes were plotted against f_2 for every level of the primary tones, thereby defining the so-called DP-gram. This term was proposed because of the good correspondence between plots of DPOAEs as a function of frequency f_2 and pure-tone audiograms when hearing losses due to OHC damage are large.

A global analysis was performed on the Békésy tracking results and the DPOAEs in order to detect the significantly abnormal ears. For this purpose, a limit between normal and abnormal ears was set according to a normative data base that we constructed using age-matched adult subjects unexposed to occupational noise. This limit was (1) hearing threshold > 20 dB HL for the Békésy audiogram, and (2) DPOAEs below the noise floor at 60 dB SPL primary levels at the frequency being tested. We deliberately focused this study upon ears with hearing losses ≤ 40 dB that had no hearing loss at the previous audiometric control performed 2 years previously. Forty-two ears were available for analysis; 33 had an abnormal audiogram. The percentage of false negative cases was calculated, considering the audiogram as a "golden standard." Nine ears had a normal audiogram, although the contralateral ear was impaired. In these cases, the study aimed to detect possible

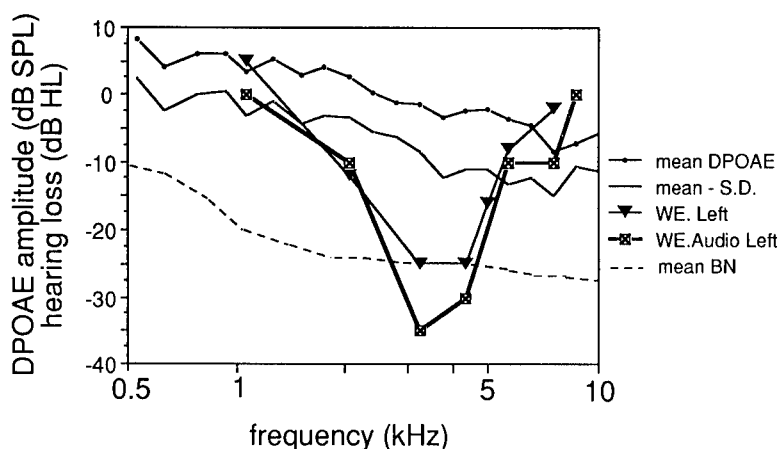


Figure 7-2 (◻) Pure-tone auditory thresholds and (▼) DPOAEs are plotted with the same scales for subject WE (left ear) suffering from NIHL. Pure-tone auditory thresholds were derived from Békésy sweeps as the median levels between inversions in the tracking procedure, in a 500 Hz wide interval around the frequencies of interest. For DPOAEs, equilevel primary tones at 60 dB SPL with $f_2/f_1 = 1.22$ were used. On the x axis, frequency corresponds to primary tone f_2 . The mean DPOAE amplitudes found in a control population serve as references. Mean -1 SD is also indicated. When a DPOAE amplitude is less than mean -2 SD, it is considered as abnormal. (---) The mean background noise during recording sessions.

changes in DPOAEs that might have anticipated the hearing loss.

We also attempted to improve the sensitivity of the DPOAE test in all the cases with abnormal audiograms but normal DP-grams. For this purpose, the slopes of the DPOAE growth functions were computed for primary levels between 50 and 65 dB SPL, and compared to the normal values found in the control population (i.e. mean 0.6 dB/dB, SD 0.2 dB/dB). The results were analyzed separately for the two categories of ears with abnormal and normal audiograms.

For the first category, clearly abnormal DPOAE amplitudes were found at $2f_1-f_2$ when the auditory threshold was impaired at f_2 in 20 ears (61%). Such a case is shown in Figure 7-2. The thin lines represent the average amplitude and average -1 SD for the control population. The audiogram and DP-gram have very similar shapes. DPOAEs are no longer detectable when hearing loss is ≤ 30 dB at f_2 . In all these ears, the frequencies of audiometric and DP-gram notches coincided within 500 Hz. No better fit was possible owing to the chosen step of the DPOAE measure-

ments and the accuracy of the Békésy tracking procedure.

In contrast with the previous straightforward situation, 13 ears (i.e. 39%) presented a normal DP-gram, therefore they were identified as false negative cases of the DPOAE technique. This number is conspicuously large. Typical examples are plotted in Figure 7-3. Figure 7-3a shows DPOAE amplitudes larger than average in the impaired frequency range; Figure 7-3b shows a normal DP-gram. It is noteworthy that neither of these subjects presented a hearing loss of more than 35 dB HL. However, the slope of DPOAE growth function around 60 dB primary levels was larger than 1 dB/dB in eight cases (out of 13 false negative). Such abnormal slopes were found when f_2 coincided with the audiometric notch. Hence, a criterion based upon slope analysis might be proposed in order to improve the sensitivity of the DPOAE test.

For the nine ears with normal pure-tone audiograms, atypical DPOAE characteristics were found in five cases. So far it is impossible to decide whether or not these anomalies represent an early state of cochlear impairment. It

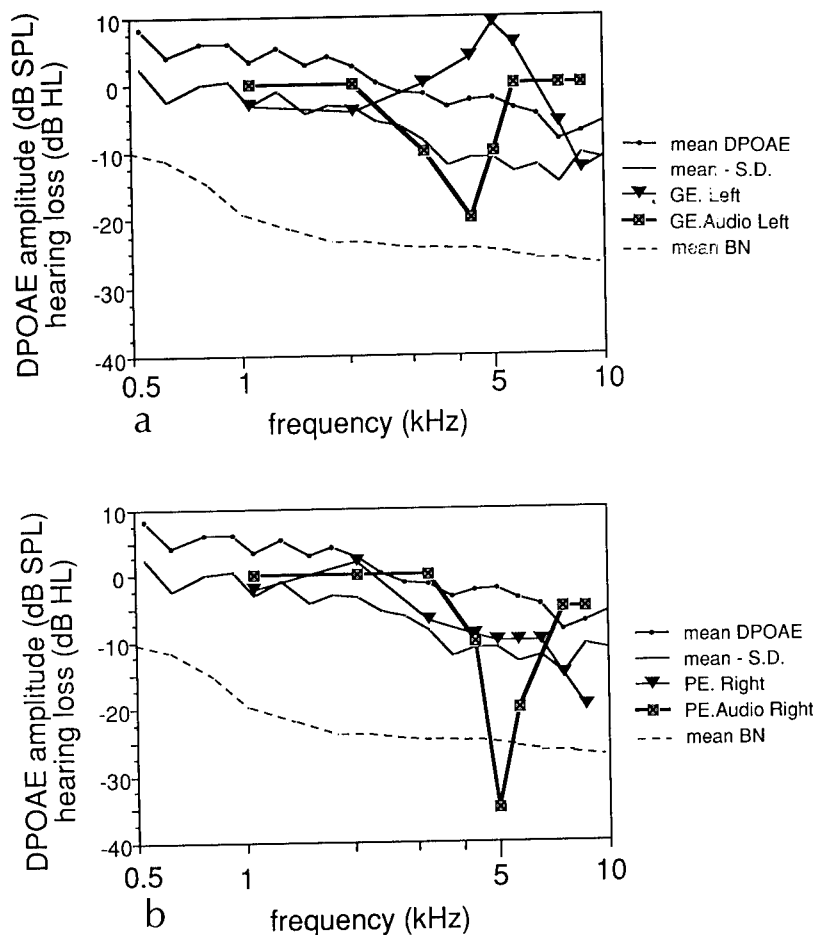


Figure 7-3 DPOAEs and pure-tone auditory thresholds are represented with the same symbol as in Figure 7-2. (a) For subject GE (left ear), DPOAE amplitudes seem larger than average at the frequencies corresponding to a small but significant notch in the audiogram. (b) Subject PE (right ear) has normal DPOAEs in spite of a narrow trough in his audiogram at 5 kHz. Note that this trough was almost undetected in conventional audiometry performed at 2, 3, 4, 6, and 8 kHz. The contralateral ear of this subject was also impaired, following an acoustic trauma that occurred 2 years earlier.

must be pointed out that the contralateral ears had clear hearing losses and that one subject complained of poor frequency discrimination in his "normal" ear in noisy backgrounds. A follow-up of these subjects will be carried out.

The small number of ears included in this study does not allow us to draw definitive conclusions as to the use of DPOAEs in the early detection of NIHLs or which characteristics of DPOAEs are most appropriate in this regard. The most frequently used DP-gram seems insufficient. As with the TEOAE, the difficulty may be the absence of a preexposure

DPOAE. The only available normal data base for analyzing the DPOAE recordings in subjects with hearing loss exhibited a large variance. Animal experiments should allow a more thorough analysis of what occurs to DPOAEs after a cochlea has been overexposed to sound.

Application of DPOAE in Animals Exposed to Loud Sounds

In an unpublished set of experiments, we exposed guinea pigs to the same type of loud

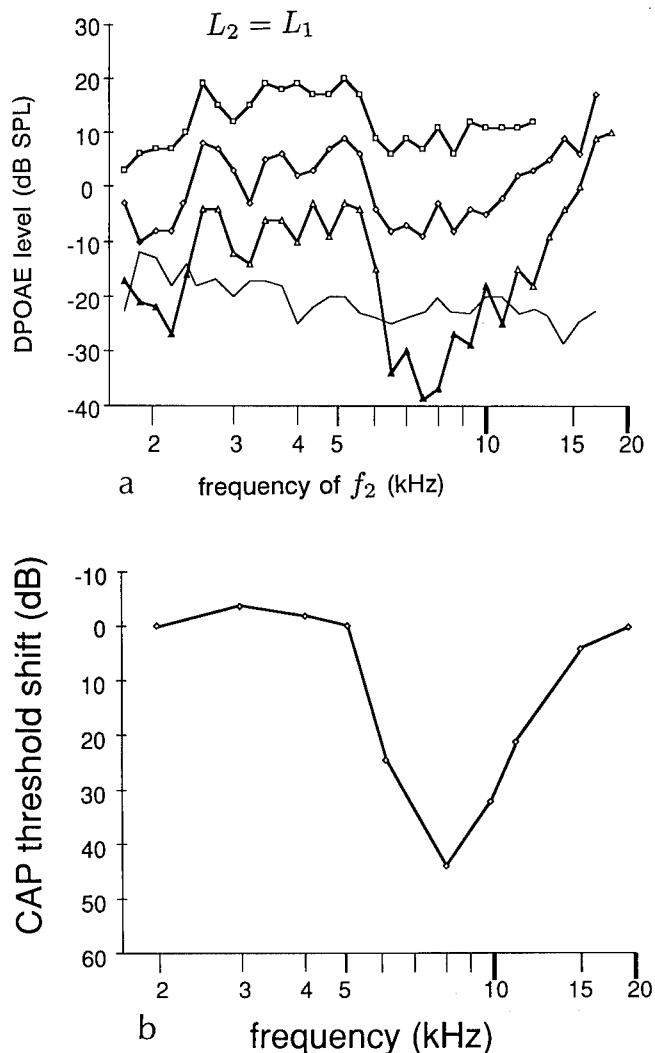


Figure 7-4(a) DPOAE in a guinea pig after exposure to a loud pure tone at 6 kHz (105 dB, 10 minutes). DPOAEs are referred to the frequency of the primary f_2 . The corresponding CAP threshold shifts are represented in **(b)**. Equilevel primary tones were used: (\square) at 70 dB SPL; (\diamond) 60 dB SPL; (\triangle) 50 dB SPL when the DPOAE is detectable, (\blacktriangle) when the DPOAE is not significantly different from background noise indicated by the thin continuous line. In spite of a 25 dB threshold shift at 6 kHz, DPOAEs were present even with 50 dB primaries. However, DPOAEs were clearly decreased with respect to their preexposure values (e.g. 22 ± 1 dB SPL between 3 and 15 kHz for 60 dB primaries).

sound exposure as described in the section on TEOAE. Figure 7-4a shows the DP-gram recorded in an ear 6 hours after it had been exposed to a 6 kHz tone at 105 dB SPL for 10 minutes. DPOAEs were elicited by equilevel primary tones delivered to ER2 earphones. f_2 was swept from 17 down to 1.5 kHz in $1/10$ octave steps and f_1 was simultaneously

varied in order to keep $f_2/f_1 = 1.20$. Three sweeps were performed with respective primary levels of 50, 60, and 70 dB SPL. Levels were controlled by a Brüel and Kjaer probe 4182 inserted within 1 mm of the eardrum. The whole experiment was controlled by a Cubdis system. The concomitant CAP "audiogram" is represented in Figure 7-4b. CAP

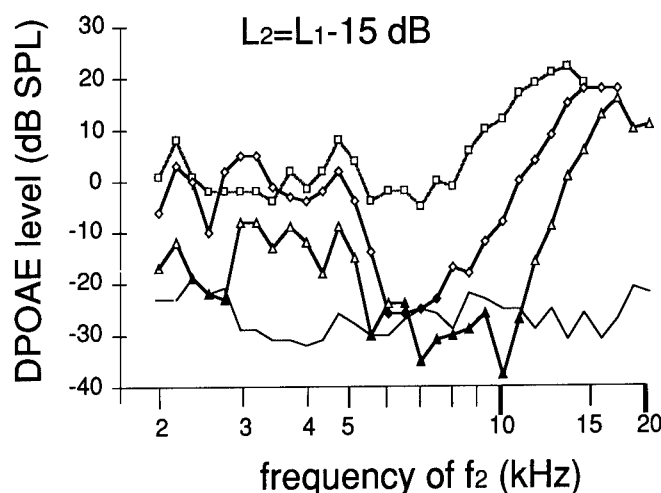


Figure 7-5 DPOAEs elicited by primary tones with levels differing by 15 dB (same symbols as in Figure 7-4; CAP thresholds are the same as shown in Figure 7-4b). The correspondence between the DP-gram and the CAP audiogram is somewhat better.

threshold shifts were found in the range 6–12 kHz with a maximum shift of 40 dB at 8 kHz.

Obviously, the DP-gram obtained with 50 dB primaries provided an accurate description of the cochlear status as described by CAP threshold changes. DP-grams obtained at higher levels were far less satisfactory. However, they were clearly altered when compared to preexposure DP-grams not shown in Figure 7-4a for the sake of clarity; their profile was very simple, almost flat from 3 to 15 kHz. The average DPOAE amplitude was 32 dB SPL (within 2 dB) for primaries with levels $L_1 = L_2 = 70$ dB SPL, 23 dB SPL for primaries of 60 dB SPL, and 13 dB for primaries of 50 dB SPL. Therefore, postexposure values were significantly decreased regardless of the primary level. However, this decrease could not have been detected with 70 dB primaries when comparing this DP-gram to a normal data base. Some DPOAE amplitudes appeared as lower than average -2 SD for 60 dB primaries, but the audiometric notch was not accurately tracked by DPOAEs. It is remarkable that DPOAEs could be detected for $f_2 = 8$ kHz with $L_1 = L_2 = 60$ dB SPL in spite of a 40 dB CAP threshold shift.

An attempt to determine the optimal ratio of levels L_2 and L_1 was also made. Before exposure, DPOAE amplitudes obtained with the

primary levels $L_2 = L_1 - 15$ dB were slightly larger than with $L_1 = L_2$ throughout the whole measured range of frequencies. This combination proved to be quite effective in tracking the notch found in the CAP audiogram (Figure 7-5) when L_1 did not exceed 60 dB SPL. A similar finding was described by Sutton et al.⁶⁸ in humans using $L_2 = L_1 - 25$ dB. In our experiment, DPOAEs returned to normal compared to the normative data base for higher levels of L_1 , that is, 70 dB SPL. Again, it must be pointed out that their amplitude was decreased with respect to preexposure values. It is clear that the slope of the DPOAE growth function was larger than normal, and larger than in preexposure conditions, and is the reason why DPOAEs quickly appeared from the background noise to almost normal amplitudes when primary levels were increased. This finding may explain why the sensitivity of diagnosis was improved in human experiments when a criterion based upon slope measurement was added. It also indicates how difficult it is to find the most appropriate combination of parameters for DPOAE production.

The presence of detectable DPOAEs with high level primary tones in spite of large threshold shifts is not surprising.³⁹ Although their source is unknown, it is most likely pas-

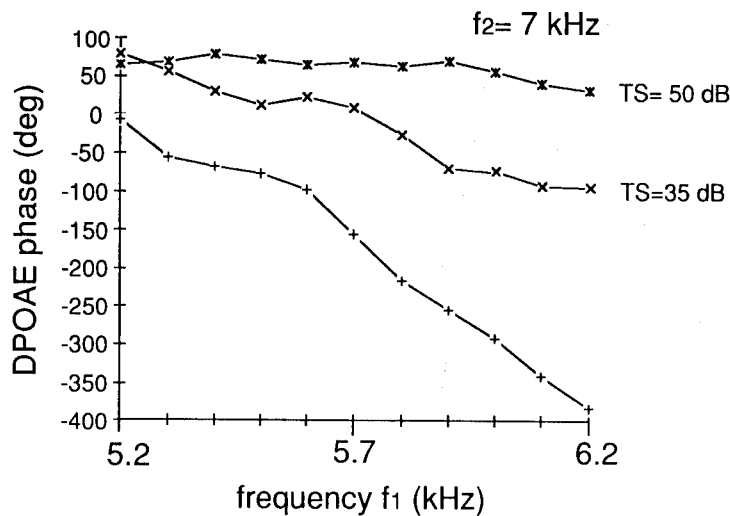


Figure 7-6 Phase variations of DPOAE at $2f_1 - f_2$ when f_1 was varied, with f_2 being kept constant at 7 kHz. Equilevel primary tones at 70 dB SPL were used. The steepest plot corresponds to the preexposure condition in a normal guinea pig and leads to a group delay of about 0.7 millisecond. Note that the DPOAE phase is plotted against f_1 , whereas group delay is given by the derivative of DPOAE phase with respect to $2f_1 - f_2$. The guinea pig was then exposed to a pure tone at 5 kHz, 105 dB SPL for 10 minutes. One hour after exposure, its threshold shift at 7 kHz was 50 dB. DPOAEs remained present with 70 dB primaries, but their group delay was close to 0. Therefore, the DPOAEs likely arose from passive untuned mechanisms. Eight hours later, some recovery had taken place with TS = 35 dB. The group delay increased to 0.25 millisecond, suggesting that the DPOAEs arose again from some tuned mechanism.

sive because OHC activity should be too impaired to take part in their generation. DPOAE phase analysis reveals this conclusion in an even clearer fashion. Figure 7-6 shows how the DPOAE phase varied with $2f_1 - f_2$ when f_2 was kept constant at 7 kHz, for 70 dB equilevel primaries. Prior to exposure, the resulting group delay was large ($t = 0.7$ milliseconds). Note that the longer group delays can be found for lower level primaries. Immediately after an overexposure giving rise to a 50 dB threshold shift at 7 kHz, DPOAEs were still present for $f_2 = 7$ kHz and 70 dB primaries, but the corresponding group delay was so short that no active mechanism could be assumed to take part in the generation of this residual DPOAE. After some recovery had taken place, and the threshold shift at 7 kHz was reduced to 35 dB, the group delay of the DPOAE was intermediate ($t = 0.25$ milliseconds), suggesting that both active and passive DPOAEs were now coexisting. Such an

approach may allow better identification of the nature of residual DPOAEs.

Although it seems possible to improve DPOAE analysis in order to improve the agreement with other audiometric measurements, it appears that a close correspondence between standard audiometry and DP-grams is found in cases of acute rather than chronic threshold shifts due to overexposure to sound.¹³ Other serious discrepancies between normal DPOAE amplitudes and growth functions on one hand, and damaged OHC with impaired auditory thresholds on the other hand, have been described by Canlon et al.¹² in the case of waltzing guinea pigs with progressive degeneration of OHCs during the first weeks after birth. DPOAE amplitudes can remain within normal limits, despite significant increases of auditory thresholds and percentages of damaged OHC. Subramaniam et al.¹⁹ have shown that during toughening experiments with repeated exposures to octave

band noises that induce resistance to threshold shifts, DPOAE growth functions gradually came back to within normal values after a period during which they decreased in parallel with hearing thresholds. Following histological examination the sacrificed animals exhibited significant loss of OHC despite the return to normal hearing threshold values after a few weeks. Thus DPOAE were insensitive to OHC losses in this particular case. In these studies, some cases of larger than normal DPOAEs were mentioned, recalling cases like the one described on Figure 7-3a.

The above results lead to two alternative hypotheses regarding the value of DPOAEs in evaluating the status of the cochlea. Either DPOAEs exhibit only a remote relation to OHC state, or DPOAEs present a differential sensitivity to the various kinds of OHC dysfunction. Quite possibly, the way OHCs degenerate is different following acute noise exposure, chronic NIHL, or genetic cochlear dysfunction. The resulting perturbation of cochlear function that leads to hearing loss may be different, and OHC mediated alterations in otoacoustic emissions may be different.

Conclusion

TEOAE and DPOAE amplitudes provide reliable information on the status of the cochlea when hearing loss due to OHC damage is well above 30 dB; milder NIHL is more difficult to detect. Thus, EOAEs appear as a good fast objective test with a screening limit value of 30 dB. So far, except under some experimental conditions, it is impossible to predict accurately the degree of hearing impairment, and to detect the onset of hearing loss.

A possible reason for this confusing state of the art is that the hypothetical sources of EOAEs are not yet clearly identified. For example, when using 60 dB primary tones for eliciting DPOAEs, the mechanical excitation is already widely distributed around the cochlear area coding for f_2 . Therefore, many OHC contribute to the overall signal with unknown phases. It is not straightforward to predict what occurs to DPOAE amplitudes when some of these OHCs are damaged,

except that DPOAEs eventually disappear whenever a large percentage of OHCs are impaired. Further work is required before EOAEs can be effectively used in the evaluation of NIHL. In particular, it is clear that more parameters of the emissions have to be studied and more thoroughly understood.

Acknowledgments

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Chapter 8

Physiological Correlates of Spontaneous Otoacoustic Emissions Induced by Acoustic Trauma

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Narrowband, spontaneous otoacoustic emissions (SOAEs) are frequently observed in the ear canals of humans.^{1,2} Some subjects hear these SOAE as tinnitus, but others do not; still others have tinnitus, but no SOAE.³ SOAEs have frequently been observed in humans with normal hearing and in nonhuman primates with little evidence of cochlear pathology, suggesting that this may be a common property of a healthy ear.³⁻⁷ However, there have been a number of striking cases of extremely high-level, tonal acoustic emissions from patients with cochlear impairment.⁸⁻¹¹ Although SOAEs are prevalent in humans, there have been only a few reports of SOAE in other animals.¹²⁻¹⁴ The SOAEs described in two of the reports were remarkable because of their extremely high amplitude, 40-59 dB sound pressure level (SPL), as well as the fact that they were associated with cochlear pathology. The results obtained by Zurek and Clark¹³ are particularly noteworthy because SOAEs were only observed in 2 of 21 ears from 17 chinchillas that had been exposed to intense sounds; SOAEs were never observed in 26 ears from 17 normal, unexposed chinchillas. These results suggested that acoustic overstimulation might provide a viable means of creating an animal model of SOAE in which it would be possible to investigate its anatomical and physiological mechanisms.

As a result of Kemp's^{4,15,16} pioneering work on OAEs in the late 1970s, as well as the

theoretical predictions of Gold¹⁷ in the late 1940s, there has been increasing support for the existence of a positive feedback mechanism within the cochlea that could enhance the mechanical vibrations of the cochlear partition. The discovery that outer hair cells (OHC) could elongate and contract at very high rates in response to transmembrane voltage, provided strong support for the view that OHCs provide positive or negative mechanical feedback.^{18,19} On the basis of these earlier findings, Ruggero et al.¹⁰ proposed that the abrupt transition between a normal and damaged region of the cochlea could result in excessive positive feedback resulting in an SOAE as well as narrowly tuned, evoked OAE (EOAE). According to this model, the positive feedback for the SOAE would originate from a normal region of the organ of Corti adjacent to the damaged region. In addition, the frequency of the SOAE would correspond to the resonant frequency of the normal region adjacent to the damaged area. Consistent with this view is the observation that contralateral sound stimulation can modulate SOAE, presumably by activating the medial efferent neurons of the uncrossed olivocochlear bundle that terminate on OHCs.²⁰

Although much is known about the acoustic properties of SOAE, relatively little is known about the anatomical and physiological mechanisms responsible for SOAE because they are seldom observed in animals except after

acoustic overstimulation. We provide a comprehensive description of the acoustic and physiologic properties of an SOAE that was observed in a chinchilla that had been exposed to a high level pure tone. We do not know if the SOAE existed before the exposure because the equipment for making such measurements was not available to us at the time; however, because SOAEs have not been observed in normal hearing chinchillas, we surmise that it developed after the exposure.¹³ This assumption is bolstered by the observation that 3 out of 10 ears had SOAE after the same traumatizing exposure.

Methods

Five healthy chinchillas were exposed to a 105 dB SPL tone at 2.8 kHz for 2 hours and allowed to recover approximately 8 months before their OAEs were measured. Three ears in two animals were found to have SOAE in the 4–7 kHz region and the one with the largest SOAE is the subject of the current report. OAE measurements were carried out over a period of approximately 1 year using equipment described in a recent report.²¹ OAEs were measured with a low noise microphone (Etymotic ER10B), the output of which was fed to the A/D converter located in a personal computer for further processing and analysis. Two channels of acoustic stimuli were generated digitally and presented through two separate earphones (Etymotic ER2) coupled to the microphone assembly. In most cases, the animals were placed in a restraining device²² so that they could be tested while they were awake; however, in some cases the animals were lightly anesthetized with ketamine (15–20 mg/kg, IM) to permit testing over a longer period of time.

After the emission measurements were completed, the animal was anesthetized (sodium pentobarbital, 35 mg/kg) and prepared for physiological recordings as described previously.^{23–25} The cochlear microphonics (CM) and compound action potential (CAP) were recorded from a round window electrode and recordings were obtained from single auditory-nerve fibers using glass micropipettes (20–

40 MW, 3 mol NaCl). A computer-automated threshold-tracking procedure was used to measure the frequency-threshold tuning curves and the lower boundary of the single-tone suppression area (tone duration: 50 milliseconds, 50 milliseconds off time, tone and no-tone counting intervals: 50 milliseconds). At the end of the experiment, the animal was perfused intracardially with saline followed by 3% glutaraldehyde. The cochlea was removed, placed in fixative, postfixed in osmium tetroxide, and embedded in Spurr resin for analysis²³ at a later date.

Results

OAEs

The SOAE was measured at irregular intervals over a period of approximately 1 year. During this time, the SOAE usually consisted of a single frequency component lying between 4000 and 5400 Hz (Figure 8-1a), but on a few occasions as many as five peaks were observed (Figure 8-1b). While the largest component was invariably located around 4.5 kHz, smaller spectral peaks were sometimes observed at frequencies as high as 10–12 kHz and as low as 1.7–2.1 kHz (Figure 8-1b). The amplitude of the largest peak fluctuated from day to day in the range from 25 to 35 dB SPL.

By introducing an external tone, it was possible to suppress the SOAE. Figure 8-2 shows the amplitude and frequency of the external tone needed to suppress the SOAE (4480 Hz, 31.2 dB SPL) by 3 dB. Sound levels as low as 0 dB SPL suppressed the SOAE near the tip of the suppression contour located at 5.8 kHz, approximately one-third octave above the SOAE. The high frequency slope, low frequency slope, and the Q 10 dB value for the suppression contour were approximately 225 dB/octave, 76 dB/octave, and 8.0, respectively; these results are comparable to those of single auditory-nerve fibers in the chinchilla.^{23,24} The sharp tuning of the suppression contour suggests that the emission originates from a narrowly tuned segment of the cochlear partition.

Distortion product OAEs (DPOAEs), which provide a sensitive index of cochlear pathol-

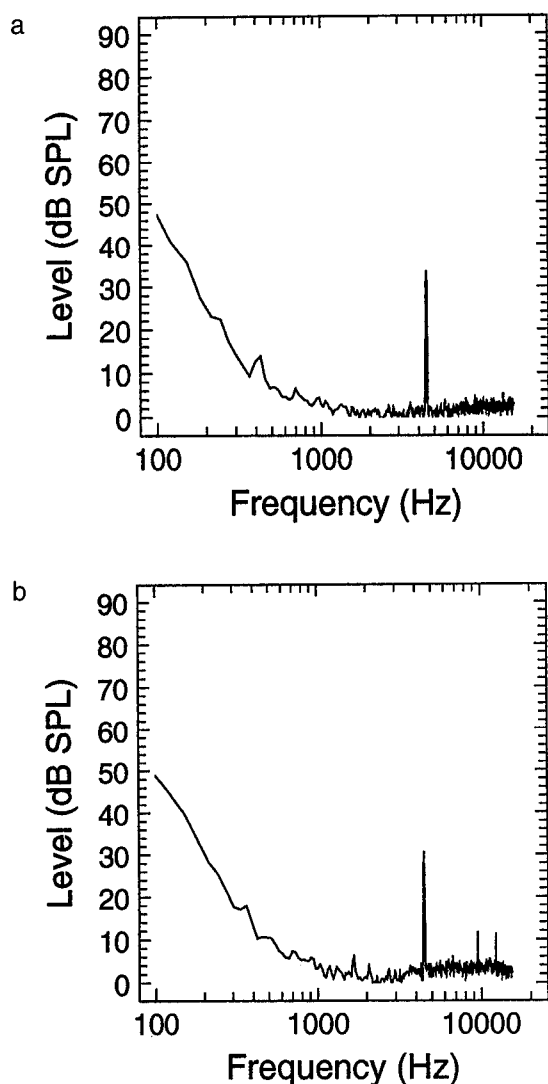


Figure 8-1 Right ear of chinchilla 3599. (a, b) Spectrum of SOAE at two different times.

ogy, were evaluated over a range of frequencies in the animal with the SOAE.²⁶ The input/output functions for the cubic difference tone ($2f_1 - f_2$) were similar to those reported for normal chinchillas²⁷; however, the amplitude of both primary tones increased in a highly nonlinear fashion for frequencies near the SOAE (Figure 8-3, left). Attenuating the f_1 (4000 Hz) input 50 dB from its maximum level of 80 dB SPL, resulted in a sound pressure level of 17 dB SPL instead of the predicted 30 dB SPL, that is, the actual sound levels were

lower than expected at input levels below 60 dB SPL. By contrast, the actual SPL in the ear canal with f_2 at 4798 Hz was slightly higher than expected when the input level was below 60 dB SPL. As expected, the input/output functions for primary tones located above and below the SOAE were linear and the amplitude of the DPOAE were within normal limits (Figure 8-3, right).

Various drugs have been shown to alter the SOAE. Sodium salicylate (350 mg/kg, IP) completely abolished the SOAE 1–2 hours after drug administration (data not shown); this corresponds to the time when salicylates reach their peak in blood serum.²⁸ The SOAE reappeared approximately 12 hours postinjection, around the time when salicylate levels fall below their half-maximum levels in the serum. Salicylates had no effect on the amplitude of the DPOAE and had only a small effect on the nonlinearity observed for primary tones located near the frequency of the SOAE. Nimodipine, an L-type Ca^{++} channel antagonist, has been reported to alter the cochlear potentials.²⁹ Nimodipine, administered orally at 1 mg/kg, had no effect on the SOAE or the DPOAE of this animal.

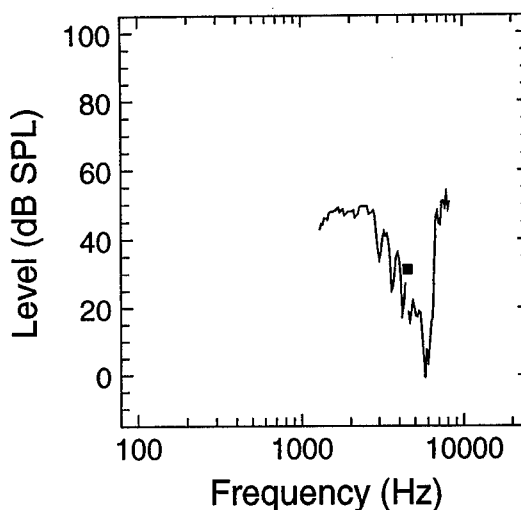
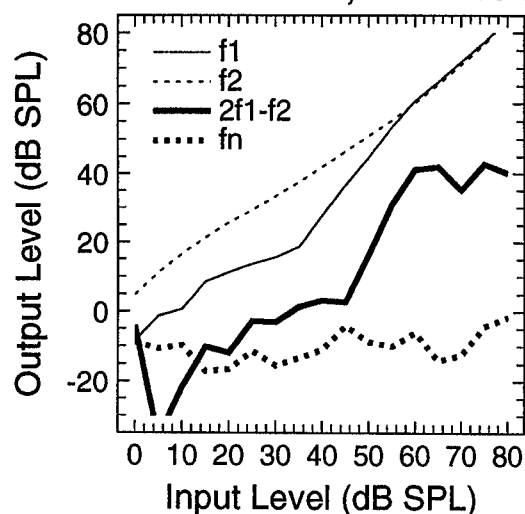


Figure 8-2 (—) Sound pressure level of an external tone needed to suppress the SOAE at 4480 Hz (31.2 dB SPL) by 3 dB. (■) Level and frequency of SOAE.

Subject 3599r, Input-Output
 $f_1=4000$ Hz, $f_2=4798$ Hz
 $2f_1-f_2=3201$ Hz, $f_n=2240$ Hz



Subject 3599r, Input-Output
 $f_1=6002$ Hz, $f_2=7201$ Hz
 $2f_1-f_2=4802$ Hz, $f_n=3360$ Hz

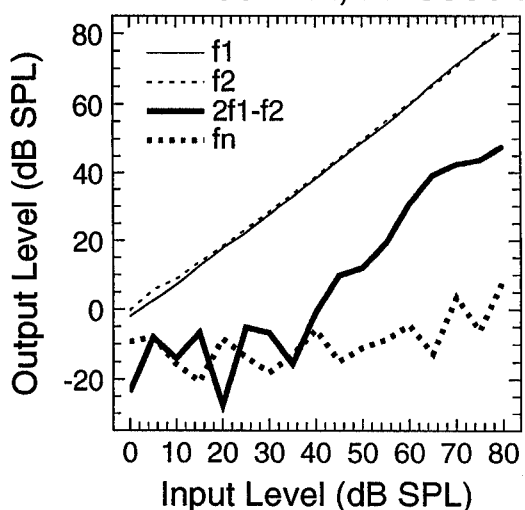


Figure 8-3 DPOAE input/output functions elicited by primary frequencies near the SOAE (top) and at frequencies above the SOAE (bottom). Note nonlinearity of the level of the primary tones near the emission frequency (top). Frequency of primary tones (f_1 , f_2), DPOAE ($2f_1 - f_2$) and noise level (f_n) indicated above each plot.

Cochlear Potentials

The CM, recorded from the round window, was monitored on an oscilloscope, audio monitor, and spectrum analyzer during the physiological experiment. When the CM was played over the audio monitor, a high pitched tone could easily be heard above the physiological noise. The amplitude of the spontaneous CM (SCM) was approximately $39 \mu\text{V rms}$. A spectrum analysis of the SCM showed a distinct peak in the spectrum near 4200 Hz; the amplitude of the peak was approximately 40 dB above the noise floor in the frequency range of interest (Figure 8-4a). A second, but much smaller peak, was also observed near 9 kHz.

To characterize the amplitude of the SCM, an external tone having a frequency slightly below that of the SCM was introduced and increased in sound level until the amplitude of the acoustically evoked CM was equal to that of the SCM (Figure 8-4b). When the level of the 3481 Hz tone was increased to 61.3 dB SPL, it completely suppressed the SCM and produced a CM that was approximately the same amplitude as the SCM. These results indicate that the spontaneous mechanical vibrations within the inner ear are actually much larger than that suggested by the SPL of the SOAE measured in the ear canal (31 dB SPL) due to the reverse transmission loss of approximately -30 dB from the cochlea to the external ear canal. The level and frequency of the external tone was varied in order to qualitatively evaluate the SCM suppression contour. Although detailed measurements were not carried out, the general shape of the suppression contour appeared to be similar to that for the SOAE.

When 1 dB of attenuation was added to the channel controlling the external tone (3481 Hz), the level of the external tone dropped to 53.9 dB SPL (a 7 dB decrease). By contrast, the level of the CM produced by the external tone was virtually unchanged (Figure 8-4c). In addition, the SCM at 4200 Hz reappeared, but at a slightly reduced amplitude (-7 dB, re: no signal condition in Figure 8-4a) along with the cubic difference tone, $2f_1 - f_2$, 2760 Hz, and the simple difference tone, $f_2 - f_1$, near 720 Hz.

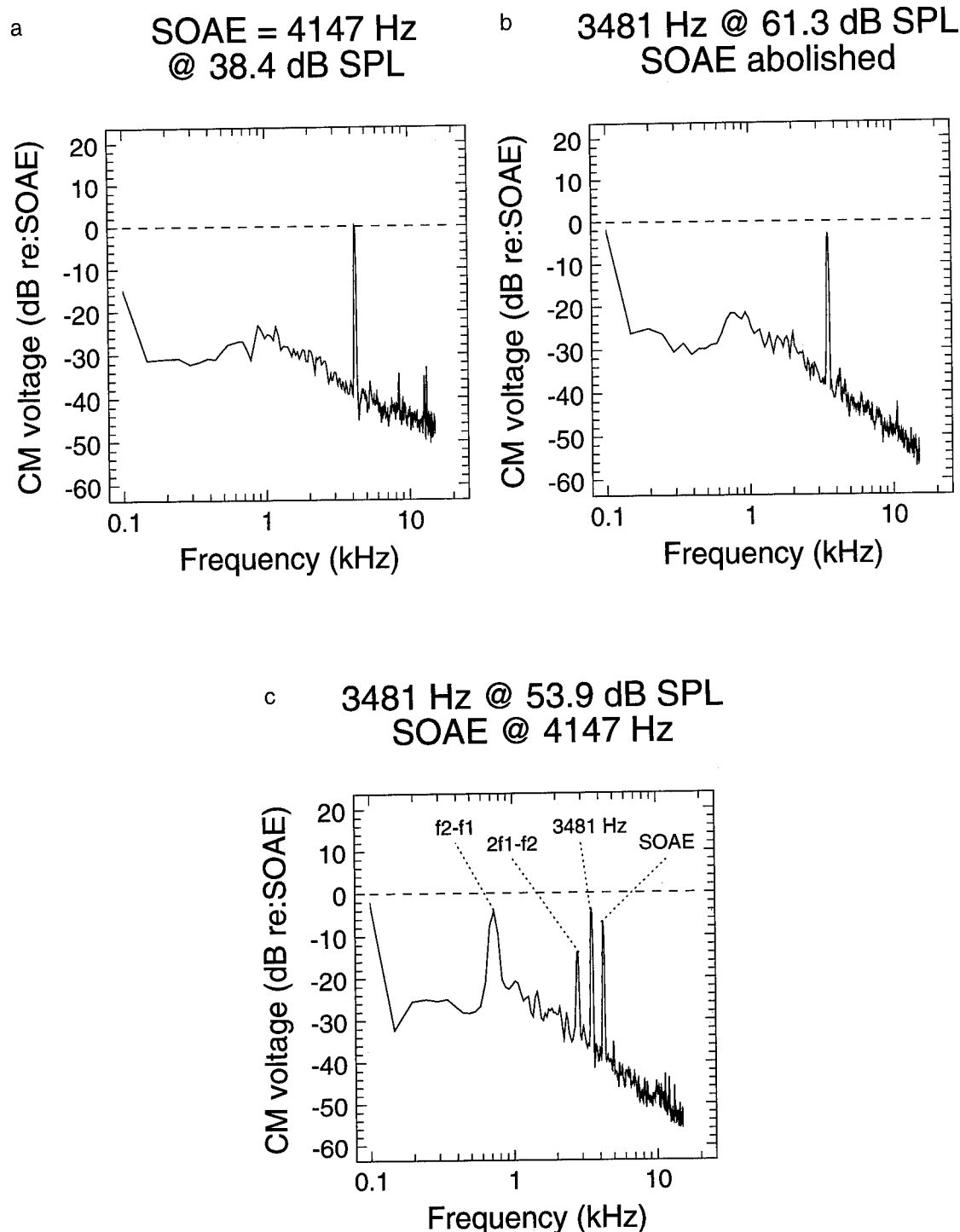


Figure 8-4 Spectrum of voltage from the output of the round window electrode. (a) Note spectral peak near 4200 Hz, in quiet, showing the SCM. 0 dB corresponds to $39 \mu\text{V rms}$. (b) External tone at 3481 Hz and 61.3 dB SPL completely suppresses the SCM and produces a CM equal to that of the SCM (see a). (c) External tone at 3481 Hz at 53.9 dB SPL partially suppresses the SCM. Note $2f_1 - f_2$ and $f_2 - f_1$ component present in the CM.

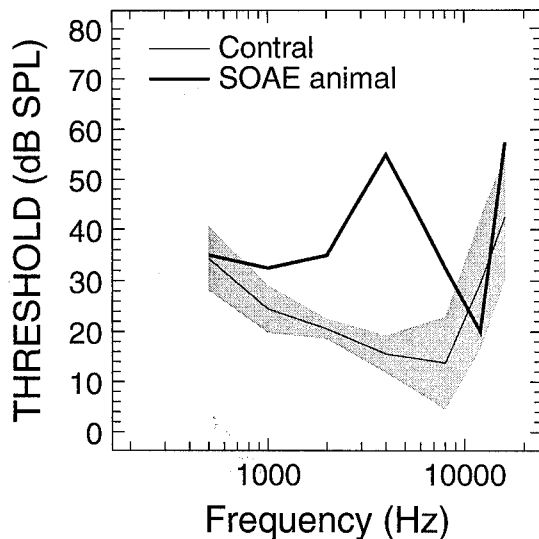


Figure 8-5 CAP audiogram from normal control animals (thin solid line: mean; shaded area: +1 standard deviation from mean). Thick solid line shows CAP audiogram from the SOAE animal.

These results show that the distortion products that are present in ear canal sound pressure are also present in the CM.

The CAP thresholds were measured in the chinchilla with the SOAE in order to determine the degree of threshold shift that had resulted from the pure-tone exposure. Figure 8-5 compares the CAP thresholds in the chinchilla with the SOAE against the CAP thresholds measured in a group of normal ($n = 21$) chinchillas. The CAP thresholds from the animal with the SOAE are elevated by as much as 38 dB at 4 kHz; the thresholds drop off rapidly at higher and lower frequencies and are essentially normal above 8 kHz and below 1 kHz. The high and low frequency slopes of the hearing loss, estimated from the CAP audiogram, are approximately 19 and 26 dB per octave. These results show that the 2.8 kHz exposure produced a punctate hearing loss with relatively steep slopes approximately one-half octave above the exposure frequency.

Single Unit Recordings

Frequency-threshold tuning curves were measured in single auditory-nerve fibers using a

computer-automated threshold-tracking algorithm.^{23,24} Figure 8-6 compares the threshold at the characteristic frequency (CF) of each tuning curve from the animal with the SOAE ($n = 100$) and from tuning curves ($n = 233$) obtained from a group of normal chinchillas. The thresholds from the animal with the SOAE were similar to those measured in normal animals for units with CFs below approximately 1500 Hz and above 8000 Hz. However, the thresholds were elevated by approximately 35–40 dB in the 4–5 kHz region and then declined rapidly above and below this region in agreement with the CAP data (Figure 8-5).

The spontaneous discharge rates of normal auditory-nerve fibers (Figure 8-7) typically range from 0 to 100 spikes/s.^{30,31} The distribution of spontaneous rates is bimodal with a narrow peak near 0 spikes/s and a second broader peak near 50 spikes/s (Figure 8-7). The spontaneous discharge rates in the chinchilla with the SOAE were considerably higher than those from normal animals. Many units had spontaneous rates above 100 spikes/s, and in a few exceptional cases the spontaneous rates

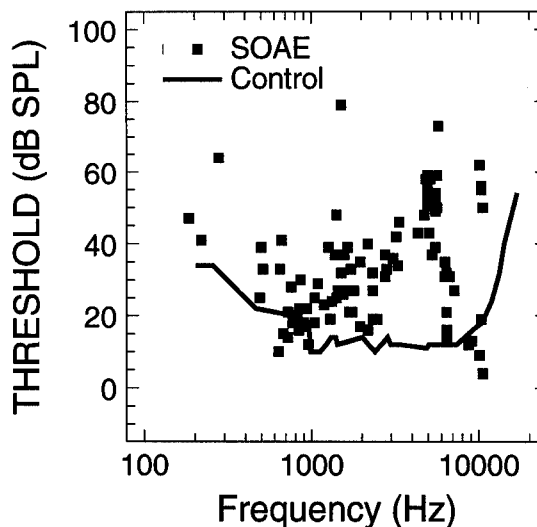


Figure 8-6 Single unit threshold as a function of CF. (—) Minimum thresholds from a group of normal control animals ($n = 233$). (■) Single unit thresholds from the animal with the SOAE ($n = 1$).

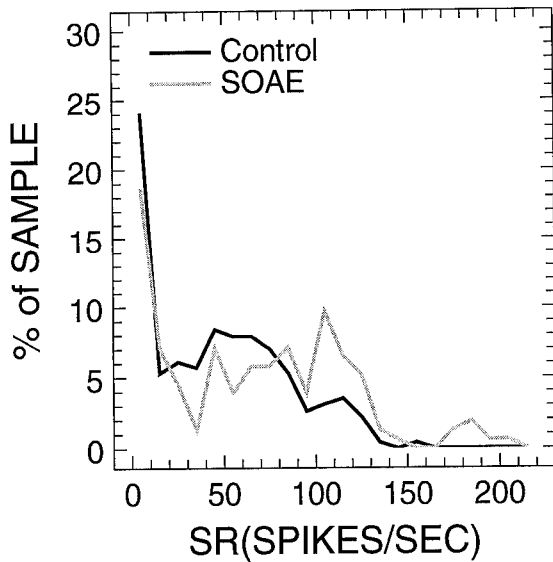


Figure 8-7 Percentage of units in sample as a function of spontaneous rates (SR) in a group of normal control animals ($n = 233$), and in the SOAE animal.

exceeded 200 spikes/s. The distribution of spontaneous rates in the animal with SOAE was also bimodal with a peak near 0 spikes/s, which was smaller than normal, and a second peak near 105 spikes/s, which was shifted to the right of the second peak seen in normal animals. The spontaneous discharge rates were also plotted as a function of CF to see if there was any variation in discharge rate along the length of the cochlear partition. The distribution of spontaneous rates in the animal with the SOAE did not show any significant discontinuity across CF except for the fact that there were few units with CFs between 2.5 and 4 kHz with spontaneous rates less than 20 spikes/s (Figure 8-8b). Moreover, the distribution of spontaneous rates across CF was similar to that seen in normal animals except for the fact that spontaneous rates were higher in the animal with the SOAE (Figure 8-8a).

The high rates of "spontaneous activity" observed in neurons with CFs near the frequency of the SOAE (4200 Hz) could conceivably represent neural activity that is "driven" by a cochlear resonator. The frequency of the SOAE would lead one to speculate that the resonator is located near the 4200 Hz region, that is, 0.58 octaves above the exposure fre-

quency. On the other hand, the tip of the suppression contour suggests the emission may originate near the 5600 Hz region of the cochlea.

The elevation of CAP thresholds in the 4–5 kHz region could be due to cochlear damage that simply elevates the thresholds of neurons with CFs in this region. However, the thresholds could also be elevated by "internal masking" of neural activity by the SOAE, that is, the SOAE could create a "line busy signal" in neurons with CFs near the frequency of the emission that would cause an increase in

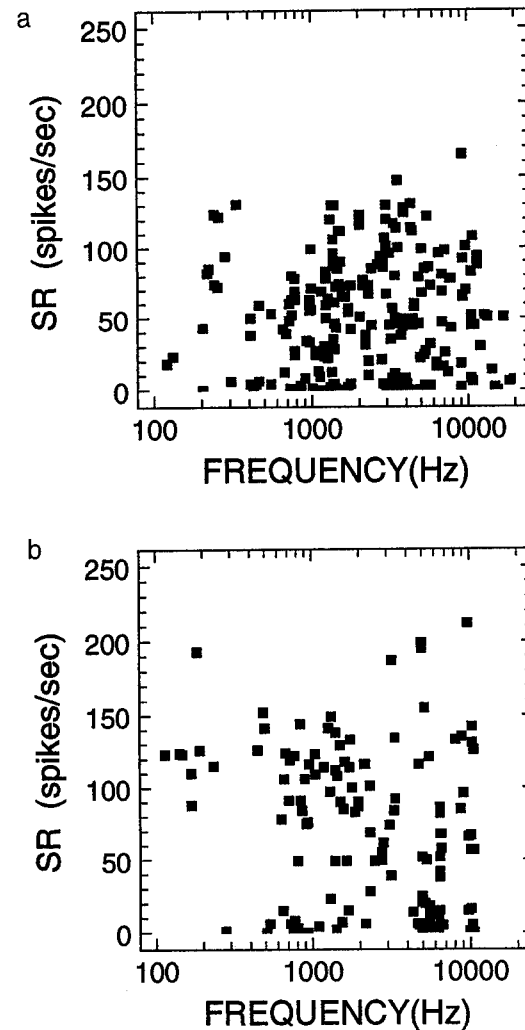


Figure 8-8 Spontaneous discharge rate as a function of CF in (a) a group of normal control animals ($n = 233$), and (b) the SOAE animals ($n = 100$).

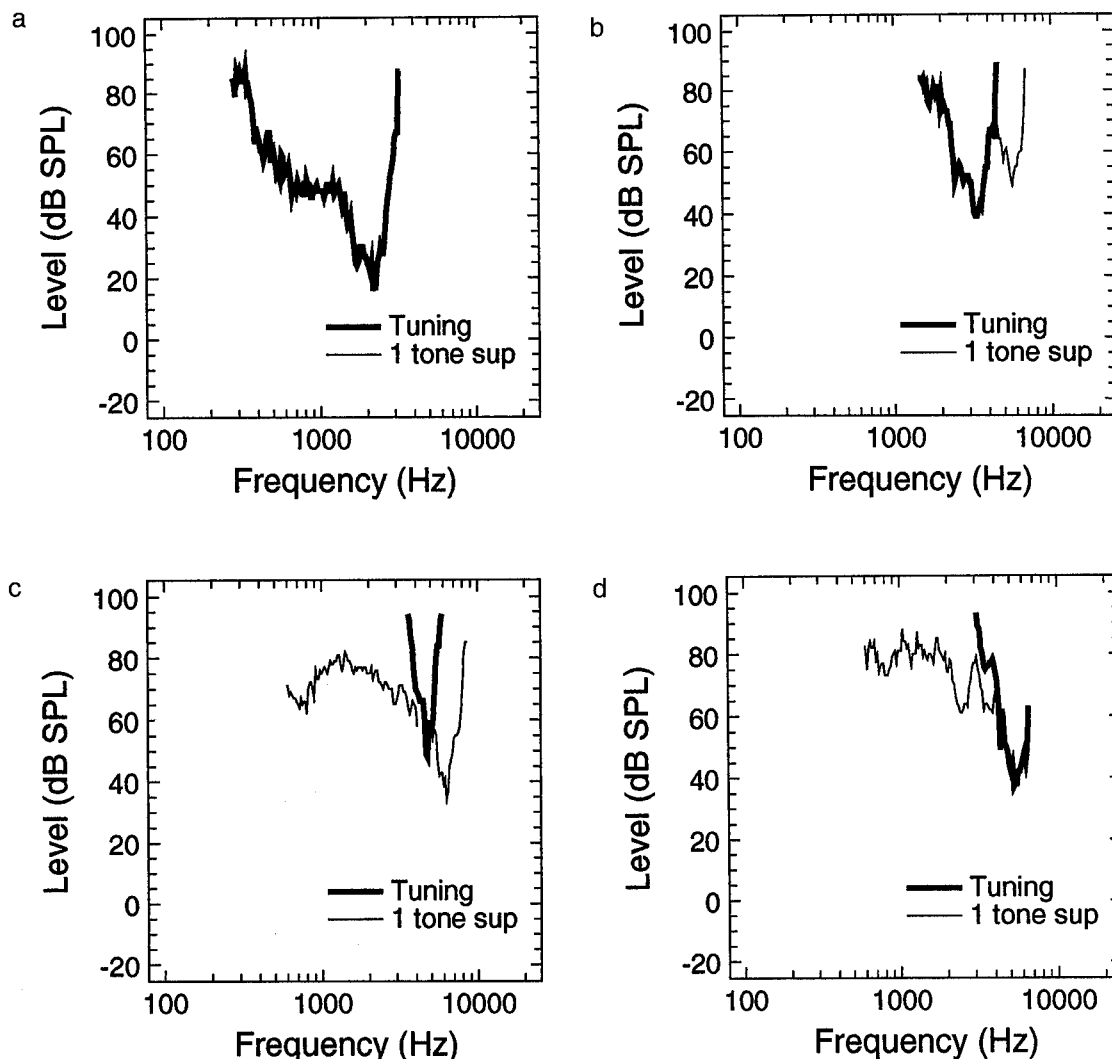


Figure 8-9 (a–d) Tuning curves (thick lines) and single tone suppression contours (thin lines) from four units in the SOAE animal. (a) 3599 U# 17, 2176 Hz 16.0 dB 116/s (b) 3597 U# 99, 3356 Hz 39.0 dB 92/s (c) 3599 U# 82, 4746 Hz 48.0 dB 116/s (d) 3599 U# 98, 5289 Hz 37.0 dB 50/s.

threshold to an external tone.^{32,33} Either or both of these mechanisms could contribute to the elevated CAP thresholds observed in this subject. To further identify the underlying processes involved, we measured both the excitatory response areas and single-tone suppression contours of auditory-nerve fibers. The thick lines in Figure 8-9 show the frequency-threshold tuning curves from four units with CFs below, near, and above the frequency of the SOAE. Units with CFs below

3 kHz had low thresholds and were sharply tuned (Figure 8-9a). The thresholds of units with CFs near 3 kHz were elevated approximately 20 dB; however, the tips of the tuning curves were still relatively narrow (Figure 8-9b). The units with the highest thresholds were located in the 4–5 kHz region. Despite the fact that the thresholds were elevated 30–40 dB, the tuning curves of many units, such as the one shown in Figure 8-9c, were quite sharply tuned. Units with CFs above 6

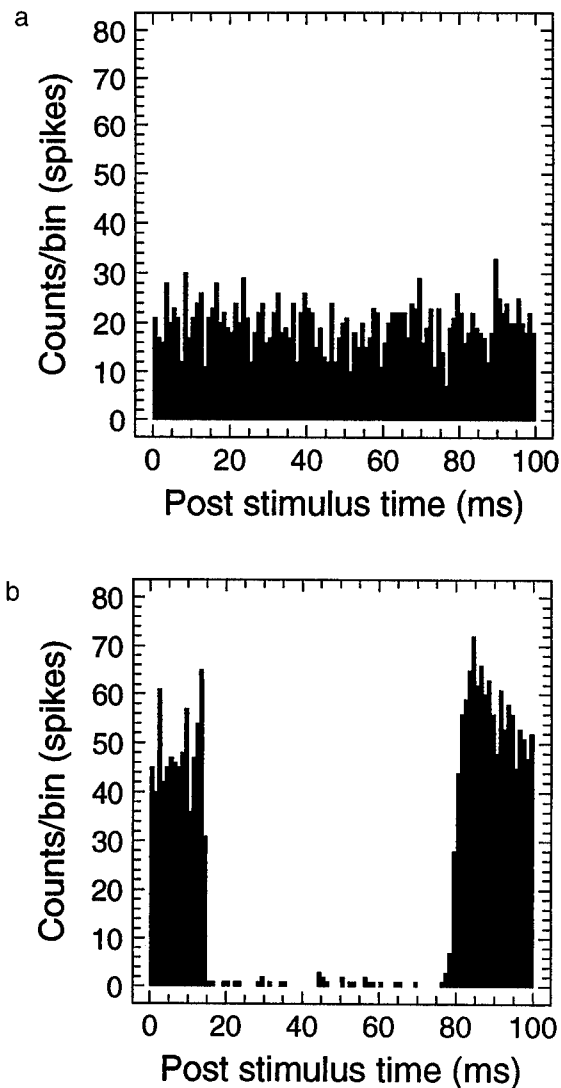
kHz had thresholds that were essentially normal and their tuning curves were as sharply tuned as those from normal animals.

The spontaneous activity of auditory-nerve fibers normally cannot be inhibited by the presentation of a single tone.³⁴ However, neural activity driven by an external tone at CF can be suppressed by a second tone above and below CF, that is, the well-known two-tone suppression effect.³⁵ If the SOAE was encoded into the spike train of auditory-nerve fibers, then the presentation of a second tone should suppress this driven response. To test this hypothesis, suppression contours were measured by sweeping a single tone through the response area in order to measure the lower boundary of the single-tone suppression contour.³⁵ Figure 8-9a-d shows the boundaries of the single-tone suppression contours measured in four units in the animal with the SOAE. Three out of four units could be strongly inhibited by an external tone. The spontaneous activity of the unit whose CF (4746) was very close to the SOAE (4200 Hz) was inhibited by tones located above and below CF (Figure 8-9c). The suppression boundary above CF had extremely steep slopes, was sharply tuned, and had a threshold approximately 10 dB lower than the threshold at the excitatory CF. The suppression boundary below CF was quite broad and the threshold for inhibition was at least 25 dB higher than the threshold at CF. The single-tone suppression boundary shown in Figure 8-9c is remarkably similar to the two-tone suppression boundary that one would obtain by presenting a tone 10–20 dB above the threshold at CF and then sweeping a second tone through the response area.^{23,24,35} If, however, the unit's CF was located slightly below the SOAE, as illustrated in Figure 8-9b, then the single tone suppression boundaries were only seen on the high-frequency side of CF. Conversely, if a unit's CF was located slightly further above the SOAE, then the single-tone suppression boundary was seen only below the CF (Figure 8-9d). Single-tone suppression could only be detected in units that had moderate to high spontaneous rates and CFs between approximately 3 and 6 kHz. Single-tone suppression

was not observed in spontaneously active units with CFs above 6 kHz and below 3 kHz (Figure 8-9a).

The time course of single-tone suppression is illustrated in the poststimulus time (PST) histograms (bin width 1 millisecond) shown in Figure 8-10 for units with a CF of 5176 Hz and a spontaneous rate of 155 spikes/s. The PST histograms were collected with 48 millisecond tone bursts (1 millisecond rise/fall time, cosine gating). Low level tones below threshold had no effect on the unit's firing rate (Figure 8-10a); however, the firing rate was almost completely suppressed at higher intensities (Figure 8-10b). The stimulus reached its peak amplitude 12.5 milliseconds after the start of the histogram (10 milliseconds stimulus delay, 1 millisecond rise time, 1.5 milliseconds acoustic delay to tympanic membrane) and the onset of suppression occurred at approximately 14 milliseconds. Thus, onset of suppression occurs approximately 1.5 milliseconds after the stimulus reaches its maximum amplitude at the tympanic membrane. The 1.5 milliseconds suppression latency includes the travel time through the middle ear and basilar membrane travel time, estimated to be about 0.6 millisecond plus a transmission delay at the hair-cell–auditory-nerve synapse, estimated to be about 0.7 millisecond. This means that suppression develops almost instantaneously once basilar membrane motion is initiated by an external stimulus. At the offset of the stimulus (61.5 milliseconds), the firing rate remained suppressed out to about 80 milliseconds and then rapidly increased out to 85 milliseconds before decreasing. These results show that the latency for the offset of suppression is much longer than the onset of suppression. The latency to the offset of suppression also increased significantly with increasing intensity.

The reduction of activity during the suppressor tone could lead to the buildup of the available pool of neurotransmitter within the inner hair cells (IHC) that in turn could lead to a higher than normal firing rate after the spontaneous mechanical oscillation "kicked in." This could explain why the firing rates observed following the suppressor tone (85–



Figures 8-10 Poststimulus time histograms from a unit in the SOAE animal (CF 5176 Hz, 58 dB SPL, SR 155 spike/s. Tone bursts presented at (a) 1294 Hz, 30 dB SPL and (b) 1294 Hz, 90 dB SPL.

90 milliseconds) exceed those observed in the prestimulus interval (0–10 milliseconds).

Discussion

In response to acoustic stimulation, OHCs produce an AC and a small DC receptor potential. OHC, examined *in vitro*, have been shown to elongate and contract at rates up to 30 kHz in response to changes in trans-

membrane voltage.^{19,36,37} These fast AC oscillations are extremely robust and occur in the absence of ATP and extracellular calcium and in the presence of agents that disrupt microtubules and contractile proteins.³⁸ It has been suggested that the oscillation of the OHC may be the stimulus for activating adjacent IHCs.³⁹ These mechanical oscillations would presumably increase the “spontaneous” discharge rate of a subgroup of neurons with CFs near 4 kHz and the spontaneous discharge rate could presumably be suppressed by an external tone.

We do not know the exact mechanism by which the SOAE, SCM, and mechanical oscillations are initiated. One possibility is that spontaneous oscillations of the stereocilia bundle modulate the transmembrane voltage resulting in OHC motion that could in turn further amplify the movement of the stereocilia bundle. The OHC stereocilia are graded in length and stiffness along the length of the cochlear partition and it has been suggested that the mass of the tectorial membrane and stiffness of the stereocilia act like a tuned resonator.⁴⁰ The tallest stereocilia in the bundle are embedded in the tectorial membrane and a change in the coupling between these two structures could alter the resonant properties of the system. The tectorial membrane, which is composed of collagens and glycosylated polypeptides,⁴¹ is extremely sensitive to its ionic environment and changes in extracellular Ca^{++} and Na^{+} could alter its structural properties.^{42,43} We have no direct evidence that the ionic environment above the hair cells was altered in our experiment; however, the extraordinarily high spontaneous rates observed across units with a wide range of CFs would be consistent with such a change. Changes in the endolymphatic potential induced by drugs⁴⁴ and electrical stimulation⁴⁵ are associated with significant changes in spontaneous rate; and the high spontaneous rates observed in the present study could be indicative of a hyperpolarization of the endolymphatic potential. Relevant to this discussion is the observation that DC current can modulate SOAE in frogs⁴⁶ and subjective tinnitus in patients.⁴⁷ The proposed hyper-

polarization of the endolymphatic potential may be a necessary, but not a sufficient condition for causing an SOAE because spontaneous emissions typically occur at specific frequencies and because salicylates, which abolish SOAE, do not modify the endolymphatic potential.⁴⁸

Another possibility is that the OHC transmembrane voltage could undergo spontaneous oscillations. Spontaneous oscillations have been observed in the membrane potential of chick hair cells. The oscillations, which arise from the interaction between voltage-gated calcium channels and calcium-activated potassium channels,^{49,50} provide avian hair cells with an electrical tuning mechanism that is believed to contribute to the regularly spaced peaks in the interspike interval histograms obtained from cochlear ganglion neurons.⁵¹ A similar electrical resonance, however, does not appear to exist in mammalian OHC.⁵²

Sodium salicylate completely abolished the SOAE 2 hours after intraperitoneal administration, but by 24 hours the SOAE had completely recovered. Salicylates disrupt the subsurface cisternae, a series of flattened membranes that line the wall of the OHC plasma membrane.⁵³ This structural change is associated with loss of electromotility. Approximately 0.5 hour after removing salicylate from the bathing medium, the subsurface cisternae regain normal appearance and electromotility is restored. Collectively, these results suggest that SOAE are closely linked to the electrical motility of the OHC whereas the DPOAE may not be dependent on OHC motility because the DPOAE were unaffected by salicylate administration.

SOAE are frequently observed in human listeners and some of these subjects report hearing the SOAE as tinnitus.^{2,3} Moreover, in some cases, the tinnitus evoked by SOAE appears to be alleviated by salicylate⁵⁴ that presumably alters the electromotile response.⁵³ Paradoxically, high doses of salicylates can induce another form of tinnitus in some patients.⁵⁵

The results of the present study indicate that the SOAE are transmitted to the central

nervous system via the auditory nerve. The SOAE appear to cause an increase in firing rate among neurons with CFs near the frequency of the emission. This increase in firing rate could create a "line busy" condition that effectively masks the neurons response to external tones of a certain frequency. This masking condition could conceivably account for part of the threshold shift observed in the CAP and single unit data. The spontaneous activity induced by the SOAE can be inhibited by an external tone located above or below the neuron's CF. Interestingly, the threshold for suppressing the spontaneous activity was often much lower than the threshold for exciting the unit (Figure 8-9c). Moreover, extremely low level tones of the appropriate frequency were extremely effective in suppressing a robust SOAE (Figure 8-1). These results suggest that tinnitus maskers and aspirin could, in some cases, be extremely useful in suppressing the tinnitus caused by SOAE.^{54,56} However, some investigators have reported that masking of SOAE was possible without influencing the tinnitus and, conversely, that masking of tinnitus was possible without eliminating the SOAE.⁵⁷

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Chapter 9

Cochlear Blood Flow Changes With Short Sound Stimulation

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A variety of specific mechanisms have been proposed to account for the changes observed in the ear following high intensity sound exposure. They fall into two categories: by direct mechanical trauma to the delicate organ of Corti structures or as a result of overdriving the metabolically dependent processes of the inner ear. Clearly these categories are not exclusive or independent of one another. Compelling examples of extensive irreparable sound-induced direct damage to inner ear structures are readily available in the literature.¹ These changes can also be subtle and require detailed and sensitive measurement, an example being the depolymerization of actin filaments in stereocilia that may provide the micromechanical basis of temporary threshold shifts (TTS).^{2,3} Whether depolymerization reflects the direct effects of intense vibration on stereocilia or a compromise of a metabolically dependent biochemical process necessary to maintain their polymerization is not known at this time. TTS also occurs at sound levels too low to yield clear structural changes discernible by light or electron microscopy.⁴ Functional changes in sensitivity can also be associated with changes in non-sensory elements, such as swelling of the lateral wall cells in the stria vascularis⁵ and swelling of the afferent nerve endings about the hair cells⁶ or supporting cells of the organ of Corti.⁷ The review by Slepecky⁸ describes the mechanical changes in greater detail. Sensitivity decreases are also associated with changes in physiological variables, such as the

level of the endocochlear potential^{9,10} or hair cell receptor potentials.¹¹

In the case of metabolically dependent damage, the changes seem to reflect the inability of the cochlea to maintain an appropriate homeostatic environment in the presence of stressful stimulation. Moreover, the observation of the recovery of TTS would seem to provide strong support for the involvement of metabolically dependent processes in the etiology of noise-induced hearing loss.

The most essential characteristic of metabolic homeostasis is adequate organ blood flow. Local homeostasis will depend on the provision of adequate O₂ and nutrients and the adequate elimination of waste products. In the inner ear, the highly dependent relationship between cochlear function and cochlear blood flow (CBF) has been well and extensively documented.^{12-15b} Furthermore, a relationship of the recovery process of TTS (and extent of induced permanent threshold shift) to tissue oxygenation and CBF can be inferred from studies using oxygen respiration "treatment" in animals¹⁶ and in humans.¹⁷⁻¹⁹ These relationships urge the continued study of the change in CBF induced by loud sound.

Loud Sound and CBF: A Historical Perspective

Noise is the adequate stimulus for hearing and obviously produces energy-dependent activity in the end organ. The driven responses can

be expected to be graded with sound intensity. For example, sound leads to the redistribution of ions in cochlear fluid compartments and subsequently the generation of metabolic waste products of biochemical processes (dependent on O_2) that perform the housekeeping operations of the inner ear to recover the baseline ion distributions. These processes may demand an increase in CBF. High intensity sound, on the other hand, may have direct mechanical effects on the vasculature and consequently on CBF. Data in the literature is controversial on whether certain sound conditions increase or decrease CBF. Moreover, when sound-induced CBF change occurs, the physiological mechanisms responsible for the changes are unknown.

The role of CBF in cochlear pathology has been a long-standing issue in otology and the hearing sciences. Historically, it has been difficult to determine the relationship of CBF to cochlear function because the vascular structures are difficult to access without compromising their function. Early measurement techniques did not permit a dynamic and quantitative measure of CBF. This situation changed following application of the microsphere trapping technique, intravital microscopy observation of red cell velocity, and laser Doppler flowmetry (LDF) technology to the inner ear. Nevertheless, a significant body of literature based on histological assessment of CBF provides a framework on which to build and compare new results. It is from these histological studies, reflecting largely the work of Hawkins²⁰ and Axelsson and colleagues,^{21,22} that a strong notion of the flow-reducing effects of high-intensity sound can be gained.

Histological measures of CBF by their very nature are postmortem and static. Rather than demonstrating the dynamic process of blood flow, they indicate the state of blood flow parameters at the time of death. The fact that histology does reveal the effects of noise exposure (in experimental animals) is consistent with a powerful effect of noise exposure on CBF: the changes that occurred apparently were not obscured by the trauma of sacrifice,

causing the interruption of systemic blood pressure and organ flow.

The experiments that have examined the effects of noise on histological measures of CBF have all employed relatively high intensity exposures (both continuous and impulse noises) varying in duration (typically 30 minutes to 12 hours) and survival time following sound exposure (from seconds to 45 days). Yet, in any given study these parameters can be varied only over a relatively narrow range.

The potential for quantitative measurement of the morphology of involved vessels is an important aspect of histological studies. However, the results have been modest. The following changes have been most consistently noted. The overall pattern of the loud sound effect on red blood cells (RBCs) appears to be a decrease in the number of these cells, with some coalescing of cells into aggregations, that is, uneven hematocrit.²² Quantitatively these changes are described as a decrease in RBC columns, a variability of RBC density, an increase in RBC aggregations with interspersed plasma gaps, and the incidence of avascular channels, as originally shown by Hawkins.²⁰ The overall effect of sound on blood vessels is reflected in vessel lumen size: there is increasing irregularity, with perivascular cells frequently compressing the vessel lumen.

While studies of the effects of sound on histological measures of vascular parameters have not systematically varied sound intensity per se, by comparing studies employing impulse noise with those using continuous sound exposure, some differences related to degree of noise trauma are apparent. Impulse noise results in more significant changes than does continuous sound, and the effects of impulse noise appears to be more concentrated in the cochlear external wall.²²

Comparison of animals sacrificed immediately with those allowed to survive after the noise exposure indicates that the vascular changes persist after noise exposure for at least 45 days. For longer surviving animals, the observed differences from nonexposed control animals were concentrated in the spiral lamina rather than in the external wall.

These results suggest that sound exposure at high intensities can cause vascular disturbances and have an effect on CBF that may be long lasting. However, they tell us little about the mechanisms for these effects. The histological pattern of sound effects on CBF is arguably more variable than the effects of noise on hearing acuity. This greater variability may be the result of postmortem effects that are inherent in the histological procedures. The modest pattern of change cannot tell us whether the CBF changes are responsible for subsequent changes in hearing acuity or whether the CBF changes are secondary to structural changes in the organ of Corti. Certainly the apparent reduction in CBF that these results indicate is paradoxical. An organ that is fatigued or damaged would appear to require increased, rather than decreased blood flow to maintain viability or to repair damage. Some answers to these questions are provided by studies that measure CBF directly.

Perlman and Kimura,²³ using intravital microscopy, observed a clear increase in RBC velocity in the lateral wall of the cochlea exposed to sounds at levels above 120 dB sound pressure level (SPL). More recently Prazma et al.,²⁴ using the microsphere technique, reported a marked increase in CBF with 113 dB wide-band noise exposure. Quirk et al.²⁵ also found RBC velocity increases for sounds above 84 dB SPL. These observations appear to contradict the histology experiments and also other CBF investigations that used nearly identical conditions. Using the microspheres technique, Hultcrantz,²⁶ Hultcrantz et al.^{26a} and Angelborg et al.²⁷ reported no change in CBF with 100 dB white noise stimulation, and Morimitzu et al.,²⁸ using plethysmography, reported no CBF change. There are many experimental factors that could account for these variable results. The difficulty of applying the microsphere technique in small animals and the trauma potentially caused by opening the cochlea to observe RBC velocity are two such factors. The new technique, LDF, provides a solution to some of the problems of CBF measurement.

Using the LDF technique, we observed a decrease in CBF following exposure to relatively high levels of sound (110 dB) for 1 hour²⁹; and this is consistent with the observations of Scheibe et al.^{30,30a} also using LDF, who demonstrated a gradually decreasing CBF with continued exposure of the ear to 120 dB.

In Perlman and Kimura's²³ original intravital observations, they state that no change in RBC velocity was observed at 90 dB, a significant increase was observed at levels above 120 dB, and above 150 dB sudden stoppages of blood flow in selected vessels were noted. These observations are generally consistent with the more recent intravital microscopy studies of Quirk et al.,²⁵ who show large increases in RBC velocity at 110 dB. Sound levels above 120 dB were associated with sudden stoppages in some vessels that were immediately reversed with termination of the sound exposure. In contrast to Perlman and Kimura,²³ the Quirk et al.²⁵ study did find RBC velocity increases at lower sound levels (84 dB SPL). However, both studies show an apparent nonmonotonic relationship between sound exposure level and CBF. Such a nonmonotonic dose-response function has also been clearly found in the sound-stimulated growth and fall of CBF measured with iodoantipyrine by Ryan et al.,³¹ and this relationship exists as well in deoxyglucose (2DG) accumulation in cochlear and auditory neural tissue.³²⁻³⁴ Canlon and Schacht^{32,33} and Goodwin et al.^{33a} observed increased 2DG uptake in the cochlea with moderate (55-85 dB) intensities of sound exposure, but high intensities (100-115 dB) caused only a small increase over the control level. Goodwin et al.^{33a} and Ryan et al.³¹ also showed a generally increasing uptake of 2DG with increasing intensities of sound. 2DG increases were observed at exposures up to 105 dB in some cochlear structures. However, sound levels may not have been extended high enough to reveal the decline in metabolism evidenced in the Canlon and Schacht studies.^{32,33}

Another approach to measure sound-induced change related to CBF and metabo-

lism is fluid oxygen determination. Red cell velocity may bear a relationship to fluid oxygen tension changes in the endo- and perilymph during sound stimulation. Various researchers have reported decreases in P_{O_2} in perilymph or endolymph in the cochlea after exposure to high intensity sound stimulation.^{9,10,35-40} Nuttall et al.⁴¹ showed that short-term moderate sound exposures (<10 minutes) do not produce significant P_{O_2} decreases, and they suggest some forms of artifact to account for the decreases seen in earlier studies.^{9,10} The key feature of the P_{O_2} studies is a declining P_{O_2} , not attributable to artifact, that occurs over the long term (approximately 1 hour) for sound exposures at high sound intensities (>100 dB SPL).

Current Status of Sound-Driven CBF Change

Across these studies, although different subjects (mice in 2DG studies; guinea pigs in the P_{O_2} studies; and guinea pigs, cats, and rabbits in the blood flow studies), different parameters of exposure (broadband noise or pure tones), and different measures of CBF or related parameters were used, there is evidence to support the view that moderate intensities of sound increase CBF and high sound levels decrease CBF. The most compelling moderate sound data are the increases in iodoantipyrine-measured CBF,³¹ the increase in intracochlear oxygen,³⁷ and more indirectly the increase in cellular metabolism.³²⁻³⁴ Long (approximately 1 hour) sound exposures were used in these studies.

The mechanisms of such CBF increases (possibly driven by metabolic demand with moderate sound levels) are unknown. At high sound intensities, CBF may increase less strongly or may decrease below the control resting level. Sound-induced CBF decrease is most compellingly shown by the decline in iodoantipyrine-measured flow,³¹ the decline in LDF-measured flow,^{29,30} the decline in oxygen tension,^{37,40} and the decline in 2DG.³³⁻³⁴ The data are compelling because the changes could be tracked in time and/or dose-response functions could be constructed.

Among the many questions that require new information on sound-stimulated CBF changes is whether short duration (<10 minute) sound at various intensities causes flow change. An early flow change could reflect neural control of circulation or, at high sound levels, mechanical damage without the contaminating influence of metabolic shortfall. Radioactive microsphere^{26,27} studies failed to show CBF change after 5 minutes of sound stimulation. Scheibe et al.³⁰ suggest that a LDF-measured blood flow decreases after electronic removal of the "sound artifact" to which LDF instrumentation is sensitive.

The current study was designed to address the short sound exposure issue using moderate and relatively high intensity band limited noise (15–30 kHz) exposure on LDF-recorded CBF in the basal turn of the anesthetized guinea pig. We hypothesized that short duration sound at moderate levels would not influence CBF. We also hypothesized that high sound levels would decrease CBF. Below, evidence is given that supports the first hypothesis, but the second one was not supported. Instead we describe a flow increase (independent of sound artifact) that is caused by systemic reactions to the loud sound.

Methods

Six pigmented guinea pigs (300–380 g bwt) were anesthetized with sodium pentobarbital (15 mg/kg) and fentanyl (0.32 mg/kg), supplemented every 30 minutes with a half-dose of fentanyl. They were tracheotomized and core temperature was maintained at 38°C with a heating pad and rectal monitor. Blood pressure (BP) was measured from a carotid cannula. The basal turn of the otic capsule was exposed via a ventral approach and the preparation was positioned and fixed during recording in a heated head holder. The PF 403 probe of a PF 4000 laser-Doppler flowmeter (Perimed Co., Stockholm, Sweden) was positioned with a micromanipulator over the lateral wall bone of the promontory, following gentle removal of the mucosa. A second LDF probe was placed over the surgically exposed basilar artery. EKG monitoring electrodes were appropri-

ately placed. The surgical procedures were reviewed and approved by the University of Michigan Committee on the Use and Care of Animals.

Sound stimuli were a limited band continuous noise (15 and 30 kHz) generated by a custom-made noise generator and delivered via a B & K 1/2 in. condenser microphone working as a speaker. The ipsilateral ears of all six animals from which CBF was measured were exposed to noise with 3 minutes on and 3 minutes off at intensities from 80 to 120 dBA by 10 dB steps. All ipsilateral ears received an additional 5 minutes of noise at the 120 dBA intensity to test for sound influence with longer exposure. The contralateral ears from three of six animals were stimulated at 120 dBA for 5 minutes to test for contralateral effects. BP, CBF (flux), LDF-measured flow in the basilar artery (BF), and a measure of cochlear vascular conductance (CVC) provided by the ratio CBF:BP were continuously recorded by a custom-made computerized chart recorder and were simultaneously tape recorded on a 4-channel recorder (Racal Recorders CTD, Southampton, UK). Pulsatile flow from the cochlea and the basilar artery were measured from the averaged LDF flux signals obtained during off-line analysis from the recorded tapes. The unfiltered processed LDF flux signal (the output signal of the instrument) from the basilar artery was filtered by a digitally controlled variable bandpass filter (5–7 Hz) (Krohn-Hite Co, Avon, MA) to obtain a smooth pulsatile flow curve. The trigger signal needed for averaging was produced by a custom-made trigger generator from the BF signal. The LDF flux signals were averaged with synchronization of the pulsatile basilar artery flow. The frequency spectrum of flux signals from the cochlea and the basilar artery were obtained with a fast Fourier transform (FFT) spectrum analyzer (Model SR760, Stanford Research System, Inc., Sunnyvale, CA). The instrumental setup is shown in Figure 9-1.

Results

Figure 9-2 illustrates the observed response to 3 minute stimulations at 80, 90, 100, 110, and

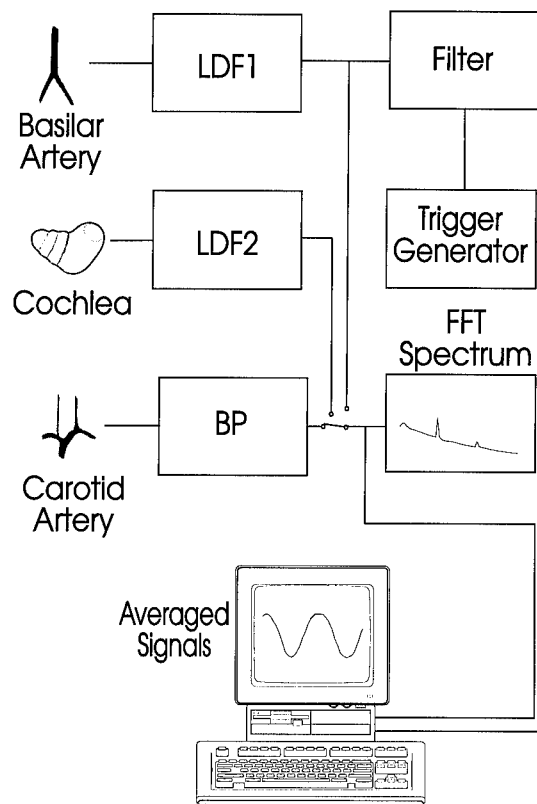


Figure 9-1 Experimental setup: The LDF laser probe for basilar artery flow (flux) is positioned outside the dura; probe tip of a second LDF was placed on the lateral wall of the first turn of the cochlea. BP signal was obtained from the cannulated carotid artery with a pressure transducer. BP, BF, and CBF were tape recorded with a computerized chart recorder and taped with an FM recorder. Averaging of the BF and CBF signals was accomplished by off-line analysis. A synchronizing signal was derived from the filtered pulsatile flux signal of the basilar artery with a trigger generator. The frequency spectrum of flux signals from the cochlea and the basilar artery were obtained with an FFT spectrum analyzer.

120 dBA, each separated by a 3 minute quiet interval. The animal in this preparation was under deep anesthesia to eliminate a contribution from systemic-evoked changes. In this case only, a slight change in BP was observed. A short transient increase in BP was noted at the onset of most noise exposures and some sound exposures produced increased variability of the BP change during and after the

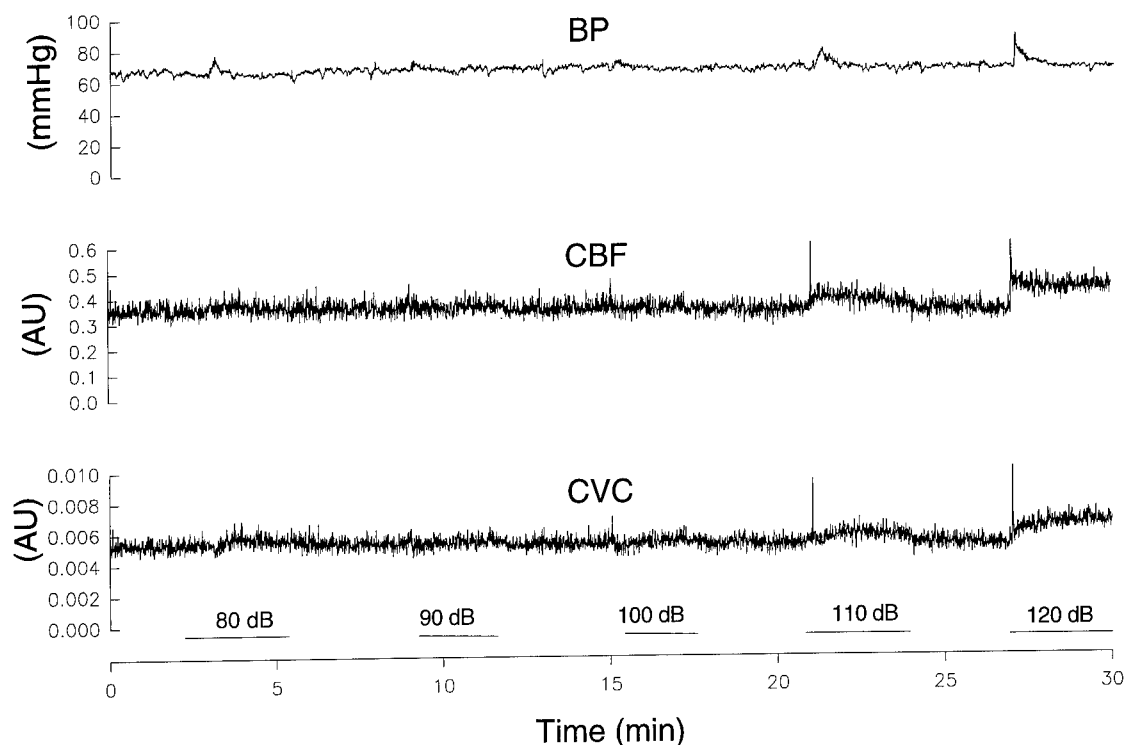


Figure 9-2 BP, CBF, and CVC response to 3 minute stimulations at 80, 90, 100, 110, and 120 dBA, each separated by a 3 minute quiet interval. There is no BP change during and after the stimulation period; however, a short transient increase in BP was noted at the onset of the loudest exposures. Little change was observed in CBF for the lower intensities of stimulation. At 110 dB there was a slow onset 20% increase and at 120 dB there was a sudden jump in CBF reflecting noise-induced artifact. Changes in CVC associated with noise exposure are clearly seen at 110 and 120 dBA in this preparation.

stimulation period (see below). Little change was observed in CBF for the lower intensities of stimulation. At 110 dB there was a slow onset 20% increase in CBF, and at 120 dB there was a sudden jump in CBF reflecting noise-induced artifacts that have been reported previously.^{30,42} The calculated changes in the conductance of the cochlear vascular bed gave the clearest response associated with the noise exposure. This response occurred above 100 dBA.

Figure 9-3a and b illustrate these same parameters measured from the ipsilateral and contralateral (respectively) cochleas in response to a 5 minute 120 dBA stimulus. This animal was lightly anesthetized and BP was therefore relatively higher than that of animals with deep anesthesia. A slow irregular oscillation of

large amplitude was noted in the BP before and after the noise exposure. The frequency of this oscillation increased and the amplitude decreased in both the ipsilateral and contralateral ear during the stimulation period at intensity of 120 dBA. The ipsilateral CBF increased by approximately 50% immediately after the onset of stimulation and maintained this level until the termination of the noise exposure. Ipsilateral CVC showed a greater increase than CBF because the mean BP decreased during the stimulation. These increases in CBF and CVC did not occur during the contralateral stimulation. The sudden onset and offset of sound-produced increase in CBF, along with the observation that an increase did not occur during the contralateral stimulation, strongly suggest that the CBF

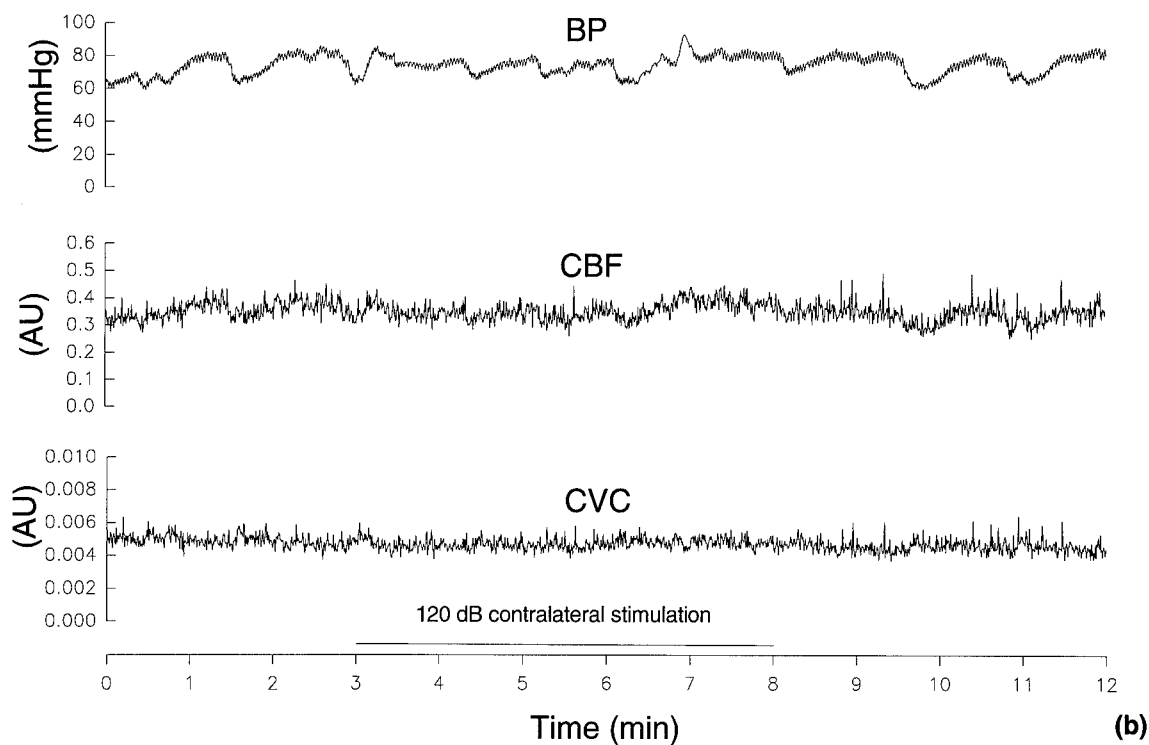
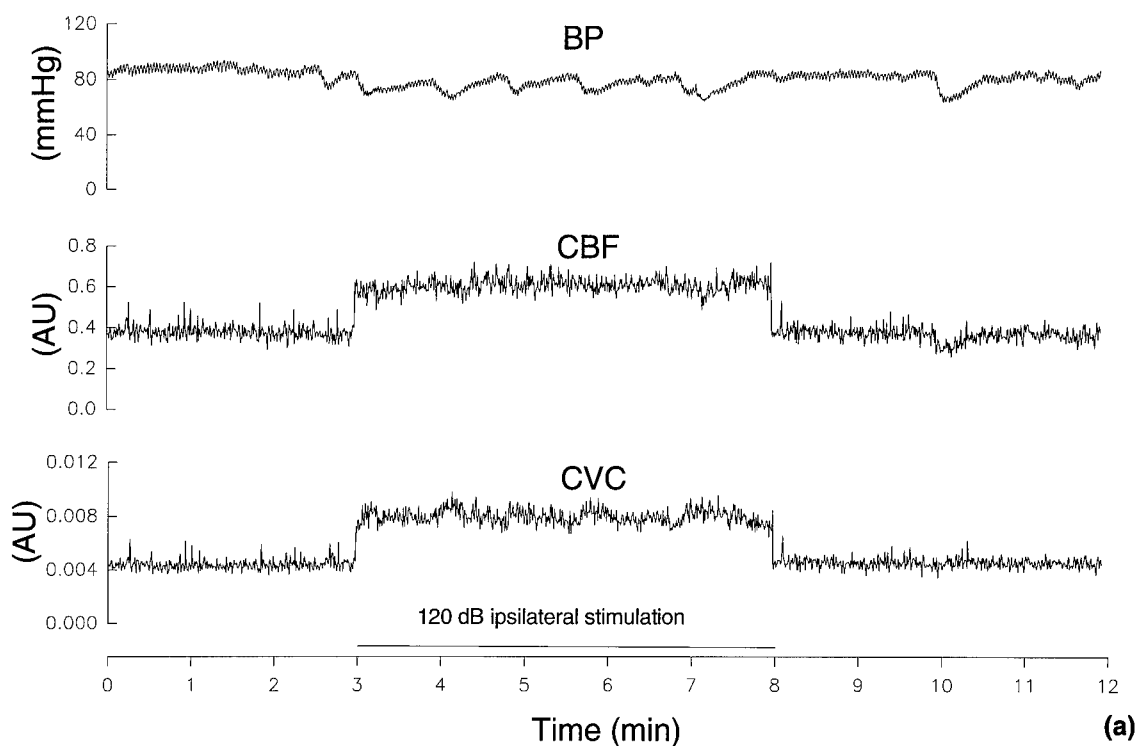


Figure 9-3 BP, CBF, and CVC responses to the ipsilateral and contralateral 120 dBA stimulus for 5 minutes. This animal was lightly anesthetized, so BP was relatively higher than for animals deeply anesthetized. **(a)** CBF increased by approximately 50% immediately after the onset of ipsilateral stimulation and maintained this level until the termination of the noise exposure. Ipsilateral CVC showed an even greater increase than CBF. **(b)** There were no changes in CBF or CVC during contralateral stimulation.

and CVC response is not a real blood flow signal but instead is a sound-produced artifact. The amplitude of sound-produced increase in CBF and CVC is dramatically variable from ear to ear.

These observations were further analyzed in two ways: the first was to examine the averaged pressure and flow response that was synchronized to the cardiac cycle and the second was to examine the frequency components of the pressure and flow responses. Figure 9-4a and b illustrate the averaged BF and CBF responses observed during quiet and during 120 dBA stimulation. Responses are shown under conditions of stimulation ipsilateral to the CBF recording. One hundred twenty-eight sweeps, triggered by every other pulsatile flow wave, contributed to the average for the last minute of each quiet and exposure condition. A clear increase in the peak to peak amplitude of the averaged CBF pulse response was observed during the 120 dBA ipsilateral noise stimulation relative to that seen in the baseline condition (BL). In addition, the averaged CBF showed a forward shift in phase that indicates a faster transmission of pulsatile flow to the cochlea during the sound exposure. The averaged BF pulse from the same animal under the same conditions and at the same time period demonstrated a smaller peak to peak increase and forward phase shift than the averaged CBF. CBF changes in amplitude and phase obviously are not proportional to BF changes. This may indicate that a local vascular response in the cochlea has occurred. However, the averaged CBF response to the 120 dBA sound stimulation is very individual, because three of six animals showed amplitude increases, one animal no change, and two had decreases. These variable CBF responses could be related to different anesthesia status.

Figure 9-5a and b illustrate rms averaged FFT spectrum of the CBF and BF responses under conditions of quiet and 120 dBA stimulation. The time domain signals from the tape were AC coupled and continuously sampled in 16.34 second sweeps. The frequency domain data were exponentially averaged 500 times with overlap of 99.8%. There are two

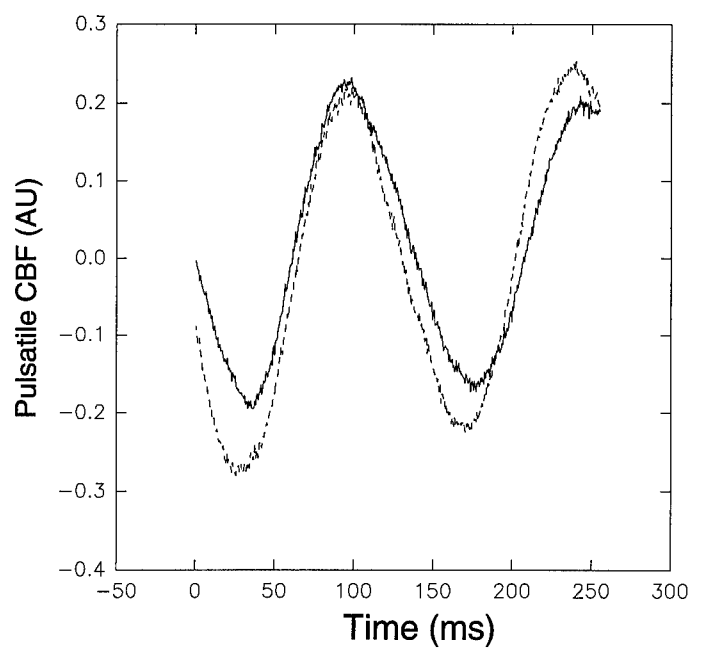
frequency peaks, approximately 7 Hz (H1) and 14 Hz (H2), in the FFT spectrum of CBF (Figure 9-5a). Peak H1 indicates the heart rate related frequency component and H2 is the second harmonic of the heart beat related peak, because $H2 = H1 \times 2$. In spite of a decrease of the low frequency component below 5 Hz, the heart rate related peak and its second harmonic significantly increased during the ipsilateral 120 dBA noise stimulation.

There are four frequency peaks in the FFT spectrum of the basilar artery pulsatile flow signal. Peak R1 is related to respiration and H1, H2, and H3 to heart beat. Peaks H2 and H3 are the second and third harmonic of the heart rate. During ipsilateral sound exposure, R1, H1, and H2 increased in amplitude and frequency, and H3 showed slightly smaller amplitude than before stimulation.

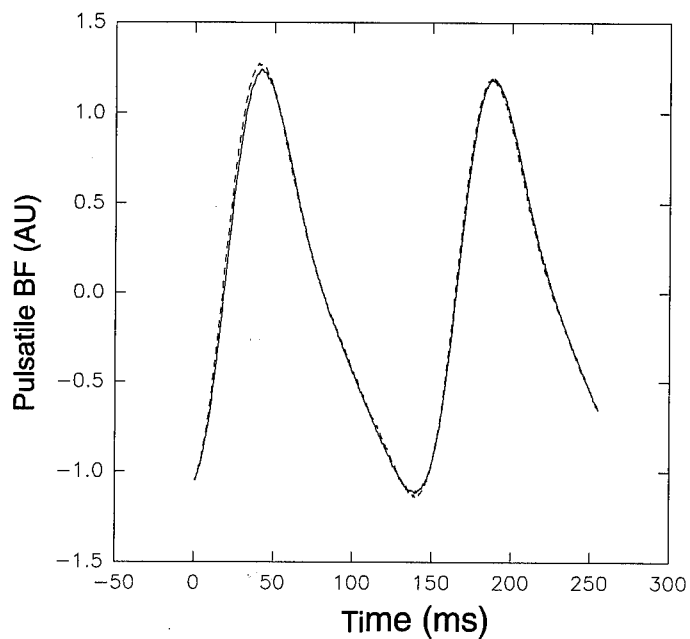
This analysis demonstrates that there are noise-evoked biological changes in the CBF. Our observations indicate that they occur at high sound levels for these short-duration exposures. The changes in pulsatile flow and energy in heart rate related peaks in the FFT can be associated with an increase in systemic BP during the noise exposure. This systemic effect may account for some of the increase in CBF. However, the data also show a direct effect of the exposure on the local vasculature of the inner ear. The variation in CBF pulse flow implies variable vascular hydraulic properties during intense sound. A similar variability does not occur in the ear contralateral to the stimulation.

Discussion

Our review of the literature suggests to us that CBF increases with moderate-intensity sound stimulation and then decreases with high-intensity sound exposure. We speculate that the mechanisms are independent of one another: that the lower level increase reflects local metabolic variables and that the high-level evoked decrease reflects systemic factors and mechanical factors that directly influence the cochlear vessel beds sufficiently to overcome metabolic factors that otherwise lead to vasodilatation. For a long-term stimulation,



— BL
 - - - 120 dBA Noise



— BL
 - - - 120 dBA Noise

Figure 9-4 The averaged (a) CBF and (b) BF responses observed during quiet (BL) and during ipsilateral 120 dBA stimulation. A clear increase in the peak to peak amplitude of the averaged CBF pulse response was observed during the 120 dBA noise stimulation relative to that seen in the baseline condition. In addition, the averaged CBF shows a forward shift in phase. The averaged BF pulse from the same animal under the same conditions demonstrates a smaller peak to peak increase and forward phase shift than for averaged CBF. CBF changes in amplitude and phase obviously are not proportional to BF changes.

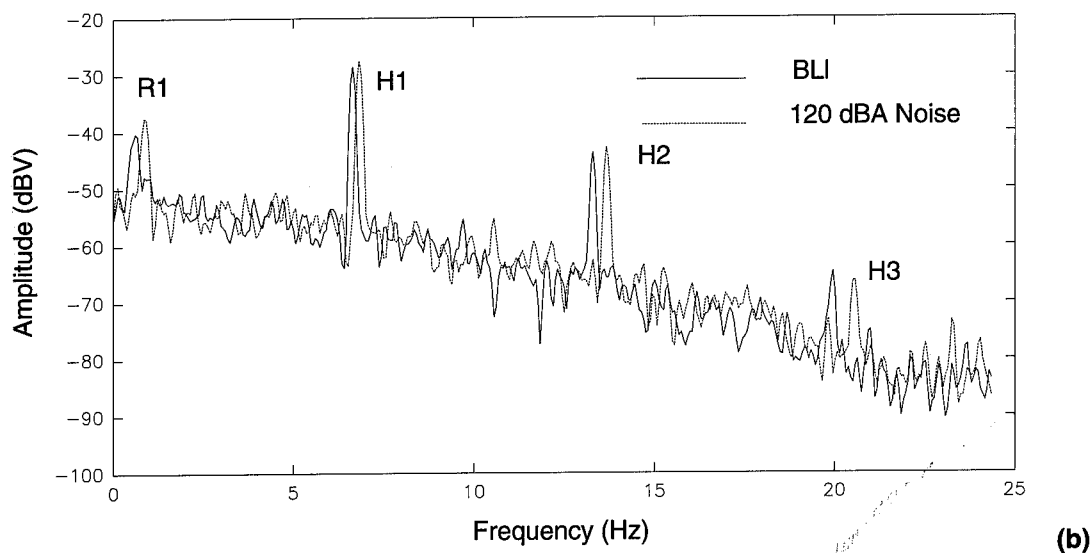
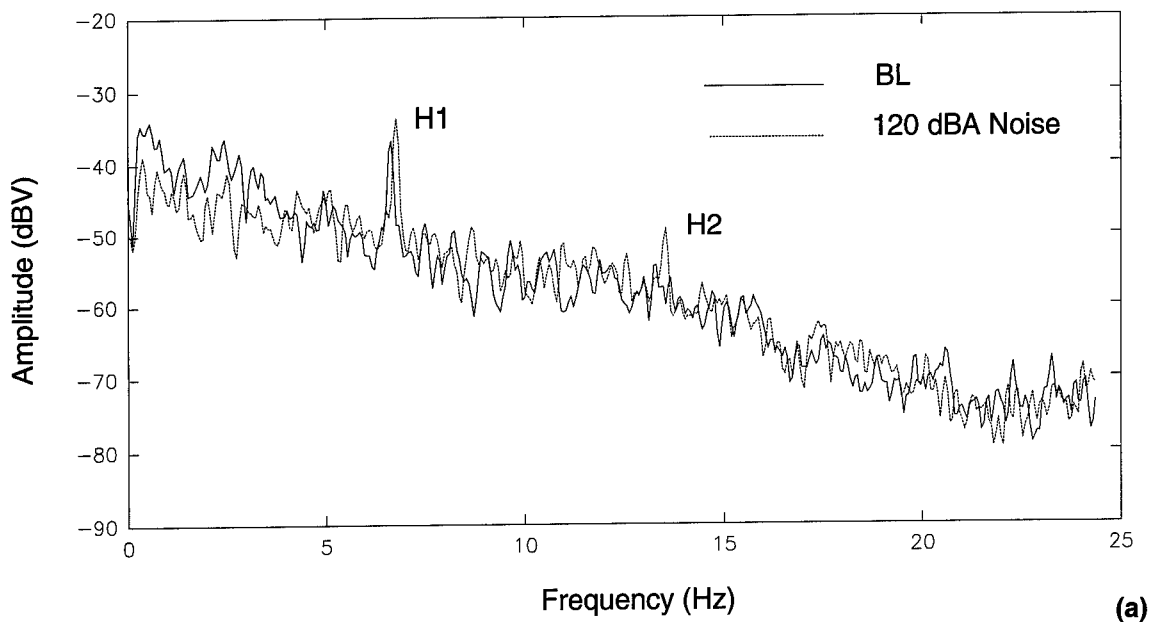


Figure 9-5 FFT spectrum of the (a) CBF and (b) BF responses under conditions of quiet (BL) and 120 dB stimulation. (a) There are two frequency peaks, approximately 7 Hz (H1) and 14 Hz (H2), in the FFT spectrum of CBF. Peak H1 indicates the heart rate related frequency component; H2 is the second harmonic of H1. H1 and its second higher frequency harmonic significantly increased during the ipsilateral noise stimulation. (b) There are four frequency peaks in the FFT spectrum of the basilar artery pulsatile flow signal. Peak R1 is related to respiration, and H1, H2, and H3 to heart beat. During ipsilateral sound exposure, R1, H1, and H2 increase in amplitude and frequency, and H3 shows a slightly smaller amplitude than before stimulation.

the data supporting this view seems clear from a variety of sources, including LDF studies,^{29,30,30a} P_{O_2} ,⁴⁰ and 2DG studies.³³⁻³⁴ In the studies using LDF, clear changes in blood flow were illustrated in the output of the LDF under conditions in which the direct mechanical noise-induced artifact, to which the LDF is sensitive, could be avoided, for example, by examining CBF immediately following offset of the stimulus²⁹ or by examining the gradual change in LDF output that might ride on a noise-induced artifact but not be influenced by it.³⁰ Such strategies are not useful in studies of the effects of short-term noise exposure and, indeed, in this area results in the literature are far less clear. As previously discussed, for moderate levels of noise (80–105 dB), many researchers^{26-28,43} observed no change in CBF. Quirk et al.²⁵ observed an increase in velocity of RBCs in the lateral wall vessels with intravital microscopy. Ryan et al.³¹ observed little or no change in flow in the lateral wall vessels with iodoantipyrine, while observing substantial increases in flow from the modiolar structures.

For high-intensity short-term stimulation, available data does not yield a clear picture of the changes induced in CBF. In the LDF study of Scheibe et al.,^{30,30a} they indicate the 125 SPL gives a decrease in the CBF. In the current study, we attempted to address this issue using LDF. Because of the known sensitivity of this instrument to acoustic energy in the cochlea, measurement and interpretation must be done with care.^{30,42} Thus, we know that with moderate- to high-intensity noise exposure, a dramatic increase in the velocity of RBCs can occur, although tissue vibration caused by the sound itself cannot be discerned in the flow of RBCs. The complex nature of "sound" artifact is illustrated by Figure 9-6 that shows a spectral analysis of the photodiode current that forms the input signal to the LDF processor. There is a clear increase in the energy distributed throughout the analysis band of frequencies. Some of this energy is in the range of the frequency band of band limited noise (15–30 kHz). However, the artifact appears as a dominant response only for high intensity stimulation (Figures 9-2, 9-3a).

In its absence, we see little change to indicate an increase in LDF-measured CBF with noise exposure between 80 and 100 dBA. This is consistent with much of the literature, including the study by Ryan et al.³¹ in which increases in CBF were observed, but not in lateral wall tissues. The LDF measurement volume is thought to be restricted to the lateral wall when a small diameter probe is used (such as that used in this study).

In an attempt to eliminate concerns for the contribution of artifact to our recordings, two analysis strategies were adopted for examining LDF-measured blood flow response with noise stimulation at moderately high (120 dBA) levels. In one case, the LDF signal (the output of the flowmeter) was averaged against a synchronizing signal derived from the pulse blood flow through the basilar artery. This site was selected, as opposed to using cardiac output or simply the EKG, because it is closer to the vessels of the inner ear, thereby reducing variability introduced by variable hydraulic properties of the intervening vessels from the heart to the measurement site in the cochlea. This procedure is similar to averaging any small signal: the noise of the measurement is reduced, permitting an extraction of the pulse flow waveform in the cochlea or basilar artery due to the cardiac cycle. To allow comparison of waveforms taken at different times, the DC value of the wave (reflecting steady flow) was subtracted, leaving only the peak to peak magnitude of the flow pulse. This approach is immune to the sound artifact that is not synchronized to the heart cycle. Analysis of the pulse waveforms of CBF before, during higher intensity of stimulation (120 dBA), showed a sound-induced increase in peak to peak amplitude and a forward phase shift (Figure 9-4a). However, these changes vary with the depth of anesthesia of our preparations.

When a similar analysis was performed of basilar artery flow, sound-produced increase in peak to peak amplitude of pulsatile flow (Figure 9-4b) was much smaller than for CBF pulsatile flow (Figure 9-4a). This unproportional increase in pulsatile CBF probably suggests that the local mechanism was involved

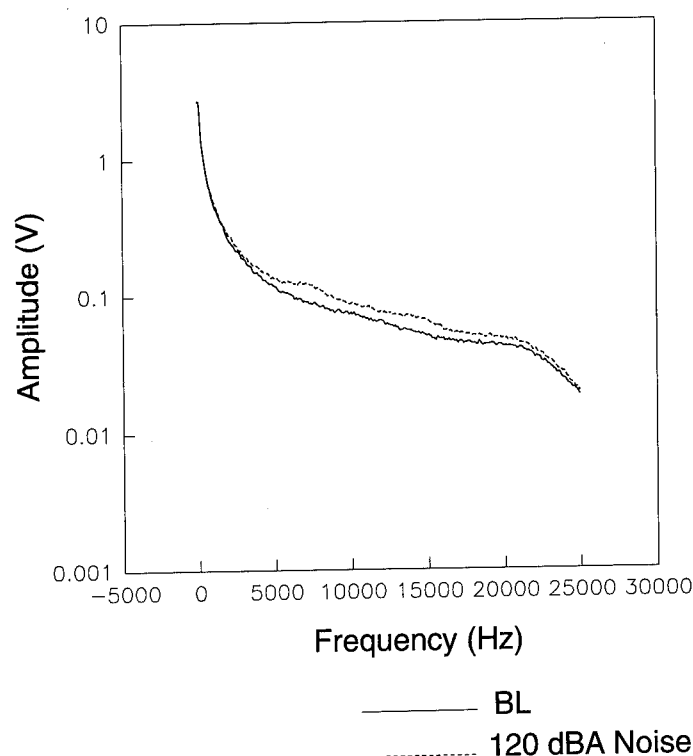


Figure 9-6 FFT spectrum of photocurrent “the unprocessed laser Doppler signal” under conditions of quiet (BL) and 120 dBA band limited sound stimulation. There is a clear increase in the energy distributed throughout the analysis band of frequencies. Some of these frequencies are in the frequency range of band limited noise (15–30 kHz).

in CBF response to short loud noise exposure. Greater increase in peak to peak amplitude may indicate an increase in pulsatile CBF and further forward phase shift could imply changes in cochlear vascular mechanic parameters during the noise exposure in this particular preparation.

The LDF signal can also be examined by the frequency domain analysis to find the frequency characteristics of the energy driving these vascular changes. Significant change in the FFT spectrum of CBF and BF was found during the noise exposure (Figure 9-5a and b). Because amplitude increases of H1 occurred in both CBF and BF, heart beat related pulsatile flow increase was mainly contributed by increased BF pulse during the noise exposure. However, amplitude increase in R1 was greater than these in H1 and H2 (Figure 9-5b),

energy of low-frequency components below 5 Hz decreased, and amplitude increase of peak H2 was greater than that of H1. Again these changes in frequency domain data for CBF are inconsistent with those for BF, indicating involvement of the local cochlear vascular mechanism in response to noise exposure. The fact that amplitude increase of peak H2 was greater than H1 (Figure 9-5a) suggests a decrease in “filter” action of the cochlear vasculature, possibly indicating an increase in vascular tone. Decreased amplitude of low frequency component below 5 Hz may be caused by changes in rhythm of vascular motor activities or/and in dynamic constriction and dilatation of arterioles. The changes, taken as a whole, lead us to conclude that CBF change in response to short-term sound exposure at high intensity levels was mainly con-

tributed by the systemic cardiovascular response, but a mild selective influence on the vasculature of the lateral wall occurred.

A number of questions remain. It is important to perform an analysis, as above, at other intensities of stimulation ranging from low level to higher levels than those examined in this investigation. It is also important for future studies to employ an LDF probe with greater separation of the fibers that will permit greater penetration and measurement of the contribution of the modiolar flow to cochlear flow during sound exposure. Based on the literature, we would expect to record significant increases in blood flow with noise exposure using a probe that analyzes deeper into the cochlea. Additional work needs to be done with pure tones to better examine the regionality of the effect. The mechanism of the CBF decrease is also not known. Nor is it known how these sound-induced changes may be modulated by other stresses and related to circulating or locally generated vasoactive agents or neural input via the sympathetic system.

Acknowledgments

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Chapter 10

Individual Differences in Peripheral Sound Transfer Function: Relationship to NIHL

Per-Anders Hellström

In a diffuse sound field, the sound pressure levels (SPL) at the human tympanic membrane are elevated compared to free field levels in the frequency range 0.2–10 kHz.¹ In the lower frequency range, 0.2–1.6 kHz, the sound transfer function (STF) is primarily influenced by diffraction around the torso; and above 1 kHz, reflections from the shoulder and the head begin to influence the STF.^{2,3} The variation in STFs at lower frequencies therefore depend primarily on the sizes and shapes of the body. Further, when the subject is seated on a chair and the sound incidence is frontal to the subject, the deviation at lower frequencies is influenced by reflections from the subject's knees.⁴ For frequencies above 1 kHz the acoustics of the outer ear are the major contributors to STF. The average STF has a peak at the $\frac{1}{3}$ -octave band around 2.5 kHz. However, from the free field to the tympanic membrane this peak can differ between individuals by one full octave from 2 to 4 kHz.¹ The major reason for these deviations is individual differences in ear canal dimensions, that is, the length, cross-sectional area, and the shape of the ear canal entrance.^{4–16} The STF from free sound field to the tympanic membrane is also influenced by the direction of sound incidence and the within subject variation in STFs could be more than 30 dB at higher frequencies.¹

These differences in STF suggest that possibility that susceptibility to noise-induced hearing loss (NIHL) may vary with the STF. That is, amplification of sounds by the outer

ear may increase the likelihood of NIHL and alter the frequency at which maximum NIHL occurs. This possibility has been supported in a study by Caiazzo and Tondorf.¹⁷ In a temporary threshold shift (TTS) experiment the subjects' ear canal lengths were artificially increased resulting in a shift from 4 to 2 kHz in maximum TTS. It has also been shown in a TTS experiment with noise exposure via earphones that the ear canal volume correlates with the frequency of maximum TTS.¹¹ Further, Rodriguez and Gerhardt¹⁸ proved in a TTS experiment with broadband exposure that the frequency of the primary ear canal standing wave was positively correlated to frequency of maximum TTS. In a subsequent experiment Hellström¹⁹ demonstrated that the STF determined the noise frequency to which subjects were most susceptible. That is, subjects with STFs that peaked in the 2 kHz range demonstrated greater TTS when exposed to noise in the 2 kHz range; subjects whose STF peaked in the 4 kHz region demonstrated greater TTS when exposed to noise in the 4 kHz region of the spectrum.

These studies support the possibility of a relationship between outer ear acoustics and NIHL. Because different independent variables were employed in the various studies, it is not immediately clear, however, whether the operative factor is STF or ear canal volume. One purpose of the current studies was to examine the relative importance of STF and gross ear canal volume in determining susceptibility to TTS. A second purpose of the cur-

rent research was to explore the possibility that permanent hearing loss is related to STF and ear canal volume as is TTS.

Methods

STF Measurement

The STFs were measured in an anechoic chamber ($3.6 \times 3.2 \times 2.0$ m) with the sound (Pink noise) delivered from a speaker (Tannoy T165) positioned in front of the subject (0° azimuth and 0° elevation) at a distance of 1.2 m. The microphone used for this purpose was a Knowles (EA 1842) miniature microphone with an attached probe. The microphone was connected to a real-time $1/3$ -octave band analyzer (Norwegian Electronic 830). The experimental method, the equipment, and its capabilities were previously described in detail.¹

Ear Canal Dimensions

The ear canal volume was measured by filling the ear canal with tempered water (37°C) from a graded (0.01 mL) 2 mL syringe. The subjects were asked to lie down with their right ear oriented upward. The tip of the syringe was placed in the ear canal entrance close to the wall. The syringe was slowly emptied so that the water could fill the canal without accumulating air bubbles. When the canal was filled to the entrance (the angled part between the canal and the pinna), the remaining water was measured and subtracted from the original amount (2 mL) in the syringe.

The ear canal length was measured by the aid of a microscope. The tympanic membrane was focused and the scale was set at zero. The ear canal entrance was then brought into focus and the difference between these two focal points was registered. This procedure was repeated until two successive measures were identical.

TTS Experiment

The subjects (32 males and 4 females) were included in the study if their pure-tone hearing threshold was better than 20 dB hearing

level (HL) in the frequency range 0.125–8 kHz. These subjects were assigned to group classifications as follows. The low-frequency groups were those whose STFs at 2 kHz were at least 3 dB greater than a 4 kHz. The high-frequency groups were those whose STFs at 2 kHz were at least 3 dB less than at 4 kHz. Those classified as the midfrequency group were those whose STFs at 2 and 4 kHz did not differ more than 1.5 dB.

The subject was seated on a chair in the anechoic chamber and the left ear canal was occluded with a foam earplug. A Békésy audiogram (0.5–8 kHz) with linear-frequency sweep was registered in the free field. The experimental sound [either 2 or 4 kHz narrow bandpass filtered (3% width) white noise] was initiated. The SPL at a position corresponding to the center of the subject's head was 97 dB, re: 20 μPa . After 10 minutes the sound was terminated, and 1 minute later a second audiogram was recorded. All subjects were exposed twice for each type of experimental sounds in a random order with more than 24 hours between exposures.

Hearing Threshold and STF Experiment

Fifty-five 17-year-old male subjects were selected for the study. All were currently enrolled in high school and were defined as the young group. Thirty male subjects employed by a road construction company and 21 musicians were also selected for this study, and labeled as the older group (age 20–60 years, mean = 38). Subjects with known ear problems or with other medical problems that could affect the results were excluded.

The subjects were tested in a soundproof room. A headset with earphones (TDH-39 with MX41AR) was placed over both ears by the experimental assistant. A Békésy audiogram (0.25–8 kHz) with linear-frequency sweep (20 Hz steps) was registered for both left and right ears. All subjects' STFs as well as ear canal dimensions for left and right ears were measured by the methods described. The subjects were divided into three groups (low-, mid-, or high-freq groups) as described.

Results and Discussion

TTS Experiment

The average ear canal volume was 1.29 mL (SD = 0.247) in the low-freq group, 0.99 mL (SD = 0.263) in the mid-freq group, and 0.88 mL (SD = 0.205) in the high-freq group. The ANOVA indicated significant differences in ear canal volume between groups [$F(13.202/2) = P < 0.0001$]. Linear regression analysis of subjects' STF and their ear canal volumes indicated a significant positive correlation at the 2 kHz $\frac{1}{3}$ -octave band, ($F = 25.113, r = 0.65$) = $p < 0.001$; and significant negative correlations at the 4, 5, and 6.3 kHz $\frac{1}{3}$ -octave bands: ($F = 4.614, r = -0.35$) = $p < 0.05$, ($F = 4.875, r = -0.35$) = $p < 0.05$, and ($F = 4.928, r = -0.36$) = $p < 0.05$.

The linear regression analysis of subjects' TTS after 2 kHz exposure and their ear canal volumes showed significant positive correlation at the 4–5 kHz frequency range, ($F = 4.176, r = 0.33$) = $p < 0.05$; at the 6–7 kHz frequency range, ($F = 4.529, r = 0.34$) = $p < 0.05$; and at the 2–8 kHz frequency range, ($F = 4.831, r = 0.35$) = $p < 0.05$. After the 4 kHz exposure the regression analysis showed significant negative correlation between TTS at the 3–3.5 kHz frequency range and the ear canal volumes, ($F = 6.27, r = -0.39$) = $p < 0.05$.

These observations suggest that, although there is an overall positive relationship between ear canal volume and STF, it is specific to lower frequency STFs. Thus, ear canal volume alone may be less predictive of susceptibility to NIHL than a more specific STF measure. In order to examine this possibility the data were reanalyzed after grouping the subjects in three categories depending on their ear canal volumes (≤ 0.90 mL, $>0.90 \leq 1.20$ mL, >1.20 mL). This procedure resulted in 10, 15, and 11 subjects in the low-, mid-, and high-vol groups, respectively. As anticipated, the ANOVA indicated significant differences in STFs between these three groups at the 2 and 6.3 kHz $\frac{1}{3}$ -octave bands: $F(8.431/2) = p < 0.01$ and $F(4.354/2) = p < 0.05$. However, the ANOVA indicated no significant differences in TTS between these groups.

Hearing Threshold and STF Experiment

The mean ear canal volume in the two groups were 1.35 mL (SD 0.32 mL) in the young group and 1.47 mL (SD 0.33 mL) in the older group. The mean ear canal length in the two groups were 22.7 mm (SD 1.82 mm) in the young group and 23.5 mm (SD 2.2 mm) in the older group. No differences between left and right ears were noted for either group. Figure 10-1 illustrates the hearing levels as a function of frequency for the two age groups. As the figure indicates, hearing levels for the older group were considerably poorer than those for the younger group at frequencies above 3.5 kHz.

Figure 10-2 illustrates hearing level as a function of frequency for younger subjects grouped according to STF frequency. The figure shows that hearing levels were significantly separated between the low-, mid-, and high-freq groups in the frequency range 2.5–5 kHz ($p < 0.05$ or less). Ear canal volumes in these groups were also significantly different, $F = 17.89, p < 0.001$ (1.184, 1.300, 1.617). These findings indicate that current hearing level, in addition to TTS, may depend upon STF, that is, the amplification provided by

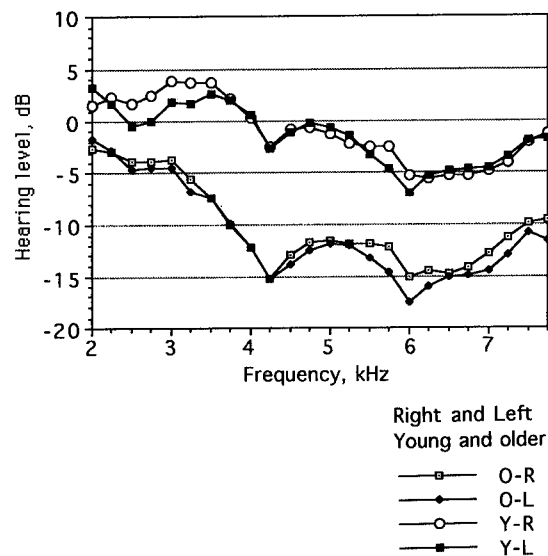


Figure 10-1 Hearing levels plotted as function of frequency for the young (Y) and the older (O) groups' left (L) and right (R) ears.

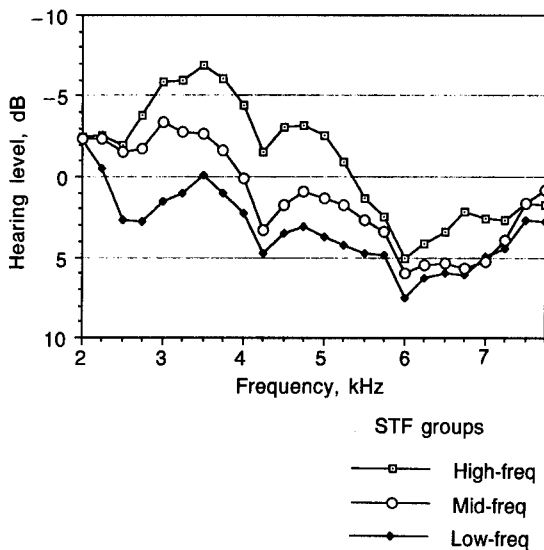


Figure 10-2 Hearing levels for young subjects grouped according to the relationship in their STF magnitudes at 2 and 4 kHz $\frac{1}{3}$ -octave bands.

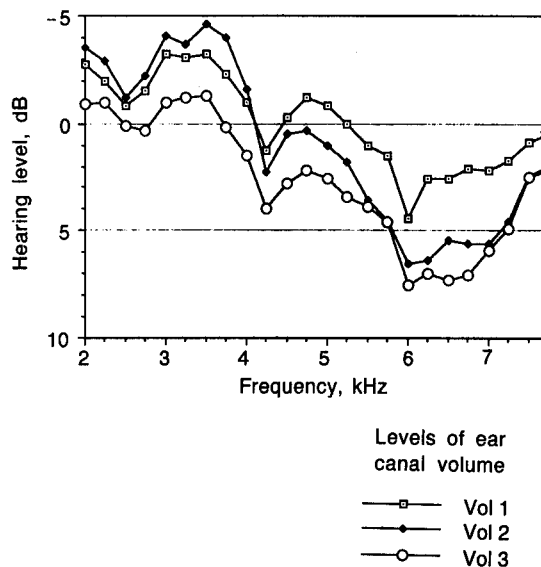


Figure 10-3 Hearing levels for young subjects grouped according to their ear canal volume. The groups are labeled Vol 1 (≤ 1.18 mL), Vol 2 ($> 1.18 \leq 1.45$ mL), and Vol 3 (< 1.45 mL).

the outer ear significantly influences hearing thresholds.

However, when the young group was reclassified according to ear canal volume, significant differences in hearing level were noted only at 6750 Hz (see Figure 10-3). In this figure the hearing levels from 2, to 7.75 kHz are plotted for three levels of ear canal volumes (≤ 1.18 , $> 1.18 \leq 1.45$, > 1.45 mL). The number of ears represented by these levels were 37, 39, and 34, respectively. The contrast between these two findings expands the notion expressed earlier that STF rather than ear canal volume may be the operative variable in predicting susceptibility to TTS. This observation suggests that STF, rather than ear canal volume, predicts hearing level.

Among subjects in the younger group, there were no differences in hearing level that could be attributed to ear canal length.

Subjects in the older group presented evidence of a somewhat different pattern of results than did the younger group. In the older group there were no significant differences in hearing levels between the low-, mid-, and high-freq groups. It is possible that the classification rules (developed for younger sub-

jects) were less applicable to the older group. Consequently more detailed examination of the older group was undertaken. Figure 10-4 plots hearing levels from the older group for three levels of 1.25 kHz STF (< 3 , $\geq 3 \leq 8$,

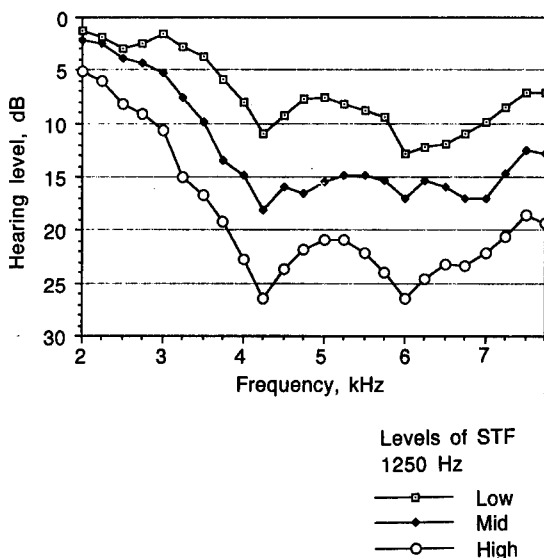


Figure 10-4 Hearing levels for older subjects grouped according to the magnitude of their 1.25 kHz STF.

>8 dB). There were 42, 39, and 21 ears at these levels. As Figure 10-4 indicates, the hearing thresholds are significantly separated in the frequency range 2.5–7.75 kHz between groups classified according to STF at 1.25 kHz ($p < 0.05$ or less). Similar results were observed when subjects were grouped according to the 1.6 kHz STF band. Significant differences were observed in hearing thresholds in the frequency range 2.5–3.0 kHz ($p < 0.05$ or less). Adopting the same type of analysis at other STF-frequency bands did not result in any significant differences in hearing thresholds.

In contrast to the younger group the older subjects evidenced a significant (all $p < 0.05$ or less) relationship between ear canal volume and hearing level in the frequency ranges 3.75–4.75 and 6.25–7.25 kHz. This relationship is illustrated in Figure 10-5 where the hearing levels from 2 to 7.75 kHz are plotted for three levels of ear canal volumes (≤ 1.20 , $>1.20 \leq 1.50$, >1.50 mL) in the older group. The number of ears at these levels were 27, 33, and 42, respectively.

As Figure 10-6 indicates, ear canal length increased significantly as a function of age ($F = 52.88$, $p < 0.001$). In addition, hearing level decreased as a function of increasing ear canal length. In Figure 10-7 the hearing levels are displayed for three levels of ear canal length (<22 , $\geq 22 \leq 25$, >25 mm) in the older group. These are significantly separated in the frequency range 2–7.75 kHz (all $p < 0.05$ or less). Whether this difference in hearing level associated with ear canal length reflects acoustic properties of the outer ear or simply difference in hearing level associated with age, remains to be determined.

As this latter point suggests, the overall different pattern of results between the younger and older subjects is difficult to interpret. These groups differ in numerous ways in addition to age. Perhaps most significant among these differences are the greater exposure to noise among older subjects and greater variability in that exposure. Nevertheless, the current results may help to explain differences in hearing level or hearing loss among older subjects.

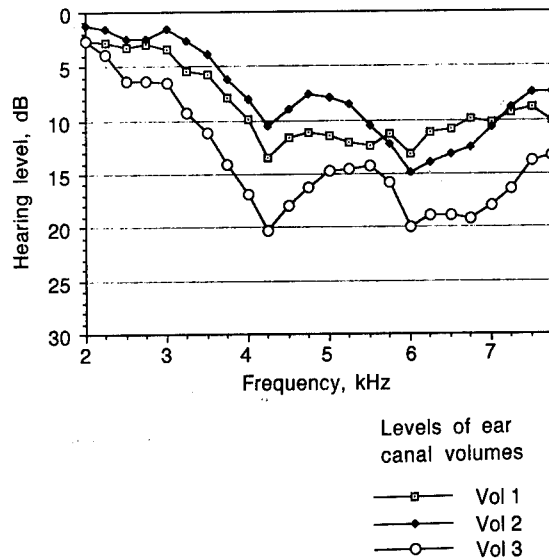


Figure 10-5 Hearing levels for older subjects grouped according to their ear canal volume. The groups are labeled Vol 1 (≤ 1.20 mL), Vol 2 ($>1.20 \leq 1.50$ mL), and Vol 3 (>1.50 mL).

In young subjects (16–18 years) there is a clear relationship between STF, spectrum of the exposure noise, and TTS. Further, there are significant correlations between STFs and hearing thresholds and some correlations between ear canal volumes and hearing thresholds. Young subjects with low-frequency dominated (2 kHz) STF magnitude have worse

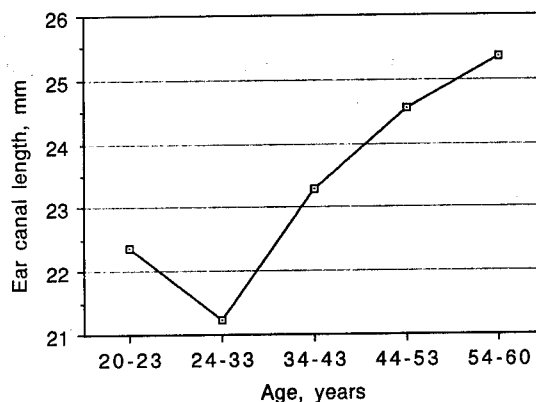


Figure 10-6 Ear canal length for older subjects as a function of age. The number of ears for each age range are 30, 14, 6, 42, and 10, respectively.

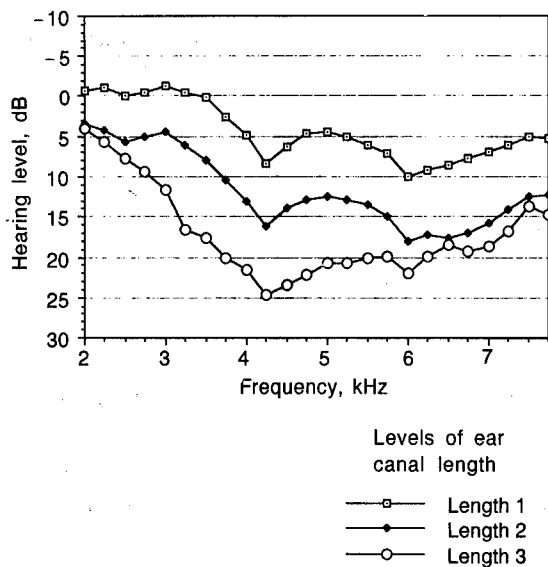


Figure 10-7 Hearing levels for older subjects grouped according to their ear canal length. The groups are labeled Length 1 (<22 mm), Length 2 ($\geq 22 \leq 25$ mm), and Length 3 (>25 mm).

hearing than those with high-frequency dominated (3–4 kHz) STF magnitudes. Older subjects with high STF magnitude in lower frequency bands than the younger subjects (1.25 instead of 2 kHz) have worse hearing than those with lower STF magnitude in this band (1.25 kHz). One possible reason for this difference in relationship between young and older subjects is the increasing ear canal length with age. The lowest frequency band where STF has a peak in magnitude corresponds to the primary standing wave in the ear canal. This frequency decreases with increasing length of the ear canal. Young subjects with low-frequency dominated STFs may belong to a group with high STF magnitudes in the 1.6 or 1.25 kHz bands when they are older. If this is true, young subjects with low-frequency dominated STFs are more susceptible to NIHL and will belong to the group with worst NIHL at older ages, if they are exposed to damaging noise levels. Why should subjects with low-frequency dominated STFs be more susceptible to NIHL than those with high-frequency dominated STFs, even if the magnitudes are comparable? One explanation is that there is

always more noise in the lower frequency bands than the higher. Most industrial noise-energy spectra show higher levels in the frequency range 1.25–2 kHz than in the 3.15–5 kHz range.

Conclusions

The results of the current studies suggest that STF is more predictive of NIHL than is ear canal volume or ear canal length per se. While these variables may permit some ability to predict hearing level or susceptibility to noise effects, it appears appropriate to conclude that they do so only insofar as they alter sound transfer functions. The current findings also indicate that STF is an important variable in predicting NIHL among younger subjects. The relationship between STF and NIHL may become less clear among older subjects for whom other variables including noise exposure history alter that relationship.

Acknowledgments

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Section //

Experimental Studies of Noise-Induced Hearing Loss

Chapter 11

Underwater Hearing and Occupational Noise Exposure

Mohammad Al-Masri and Alan Martin

In industry, occupational noise hazards to hearing are well recognized and described in many countries' legislations. Such hazards must be reduced by engineering noise control and/or personal hearing conservation programs to below acceptable limits. The maximum permissible noise exposure in air without using hearing protection varies from one country to another, but it is usually in the range 84–90 dB(A) for 8 hours.

Unfortunately, there are currently no widely accepted noise exposure limits or hearing damage risk criteria applicable for underwater use. Professional underwater divers are regularly exposed to intense noise that can reach sound pressure levels (SPLs) above 200 dB (re: 20 μ Pa).^{1,2} Underwater noise may originate from equipment operated by the divers themselves such as jet cleaning tools, rock drills, and stud guns, or from transmitted sounds such as sonar.

As in air, the effects of exposure to underwater occupational noise may result in temporary or permanent sensorineural hearing loss.^{3–7} More seriously, high exposure levels may produce vertigo, nausea, and vomiting^{4–8} that can be fatal.^{9–11}

The current noise exposure limits and hearing damage risk criteria in air use the A-weighting sound level scale (historically established from the inverse of the 40 phon equal loudness contour) that is related to hearing thresholds. However, direct transposition of these well-established limits from air to underwater use is not a simple task. This is because the hearing thresholds and hearing

mechanisms underwater are likely to be different from those in air, because the acoustic impedances of both water and the human body are the same. The most practical method of producing similar limits for underwater noise exposure would be to modify the limits in air. This would involve using the relationship between the thresholds of hearing in the two media as a correction factor to modify the A-weighting scale for underwater use. The assumption underlying this approach is that levels of equal noise exposure above the hearing thresholds in air and water will cause equal amounts of noise-induced hearing loss. This assumption is reasonable, because noise-induced hearing loss is recognized as being due to cochlear damage^{10,12,13} and the cochleae are imbedded in the temporal bone and not directly affected by immersion in water.¹⁴

Several studies on underwater hearing thresholds and mechanisms^{14–18} show that hearing thresholds are higher in water than in air (Figure 11-1) and hearing mechanisms may thus also be different. Unfortunately these studies reported a wide range of values for underwater hearing thresholds and conflicting opinions concerning the hearing mechanisms involved were proposed. These studies have been reviewed in detail by Al-Masri et al.⁹ and Al-Masri.¹⁰ The main reasons for the wide scatter of existing experimental results lies in the lack of appreciation of the significance of background noise and its masking effect on the threshold of hearing. Additionally, the subjects used in these studies may have had a hearing loss, and the majority

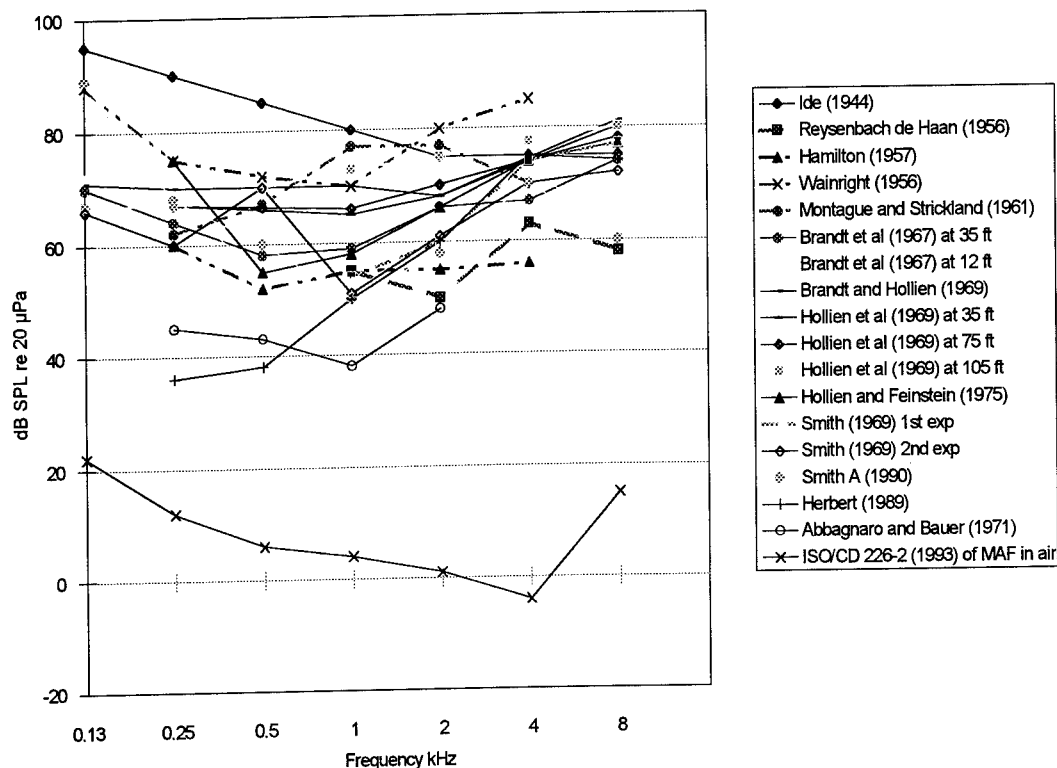


Figure 11-1 Summary of the reviewed underwater hearing threshold studies.

of these studies failed to report important information such as the audiometric hearing thresholds of the subjects, audiometric procedures, and calibration standards, as well as sound field calibration techniques. Further, the possible effects of air bubbles that may be naturally trapped in the ear canals and the effects of water depth on underwater hearing threshold have not been fully investigated.

The research reported in this chapter has three main objectives:

1. to establish a reference curve for underwater hearing thresholds over the frequency range 0.250–8 kHz;
2. to study underwater hearing mechanisms;
3. to develop a new weighting scale, equivalent to the A-weighting scale, for underwater use, so that, if applicable, the noise exposure limit and hearing damage risk criteria in air could be applied underwater in terms of the new weighting scale.

Measurement of Underwater Hearing Threshold

Methodology

To establish an underwater minimum audible field (MAF) and the relationship between the MAF in air and underwater, the experiment was divided into MAF measurements in air and underwater. The in-air measurements were conducted prior to the underwater measurements to decrease the difficulties of testing hearing thresholds underwater.^{10,19,20}

The underwater and in-air hearing thresholds of 54 normally hearing sport divers (36 male and 18 female) were tested using the procedures recommended by the British Society of Audiology.²¹ Great care was taken to minimize the level of underwater ambient noise found to be caused by both ground vibration and the transmission of airborne noise. A steel water tank (dimensions 2 × 1.5 × 1.5 m) was positioned on antivibration mounts that attenuated the ground vibration

noise transmission by at least 40 dB at all frequencies above 0.01 kHz. The ambient noise SPL in the laboratory was reduced to less than 10 dB SPL at all frequencies above 0.125 kHz. With these arrangements, the underwater ambient noise level in the tank was estimated, using a transfer function approach, to be less than 18 dB SPL at all frequencies above 0.125 kHz. This level is about 20–25 dB below the level that can be measured using available hydrophone instrumentation. The temperature of the water was maintained at 35°C for subjects' comfort. The MAF measurements in air were conducted according to the ISO 8253-2.²²

The subjects' hearing thresholds were tested, in air and underwater, using $1/3$ -octave bands of random noise because these test signals were found to provide a superior sound field uniformity compared to pure tones and FM tones. Because the naturally trapped air in the ear canals may have a significant effect on underwater hearing thresholds,^{11,19} the thresholds of 30 subjects were tested twice

with air and with air removed from the ear canals. The subjects used open circuit SCUBA, wore a T-shirt, and sat on a chair with a head rest at a distance of 1 m from the underwater loudspeaker. The loudspeaker and the head rest were 0.35 m below the water surface. The subjects held their breath while hearing thresholds were established, and great care was taken to avoid breathing-air bubble noise and other sources of extraneous noise. The test facilities and the procedures have been described in detail by Al-Masri.¹⁰

Results

The mean values of the MAF in air and underwater, with air and with air removed from the ear canals, are plotted in Figure 11-2. It can be seen from this figure that the MAF in air measured in this study is within 2 dB of ISO/CD 226-2.²³ The underwater MAF curves are frequency dependent and are significantly higher ($p < 0.05$) than the estimated underwater ambient noise level at all frequencies.

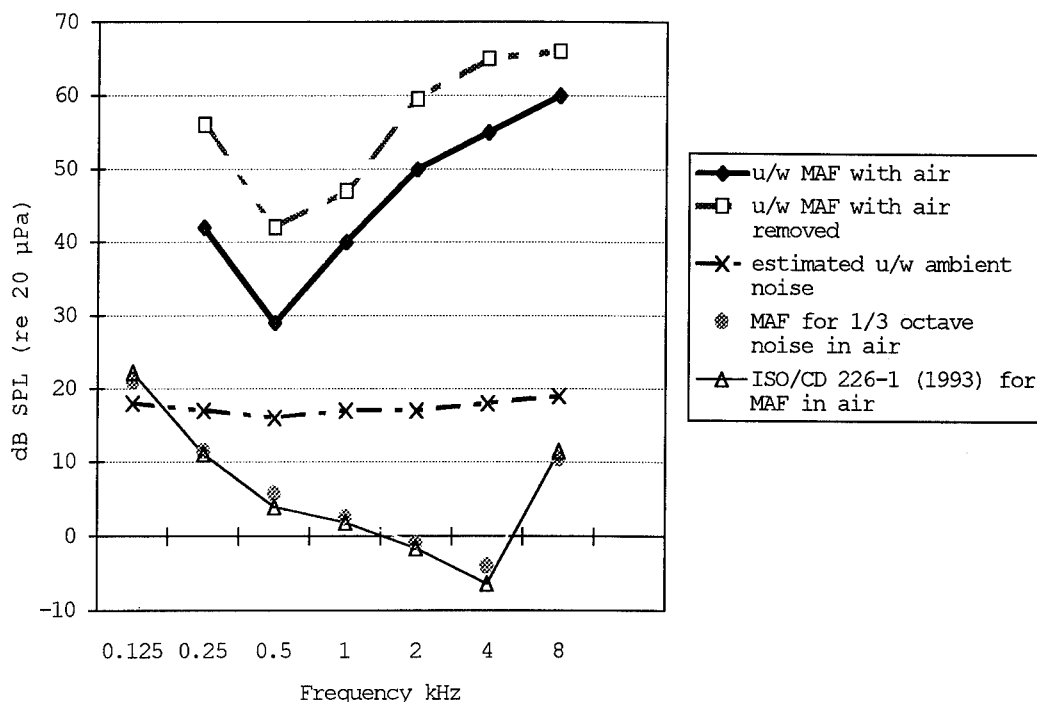


Figure 11-2 Comparison between the mean MAF values for $1/3$ -octave band noise in air and underwater.

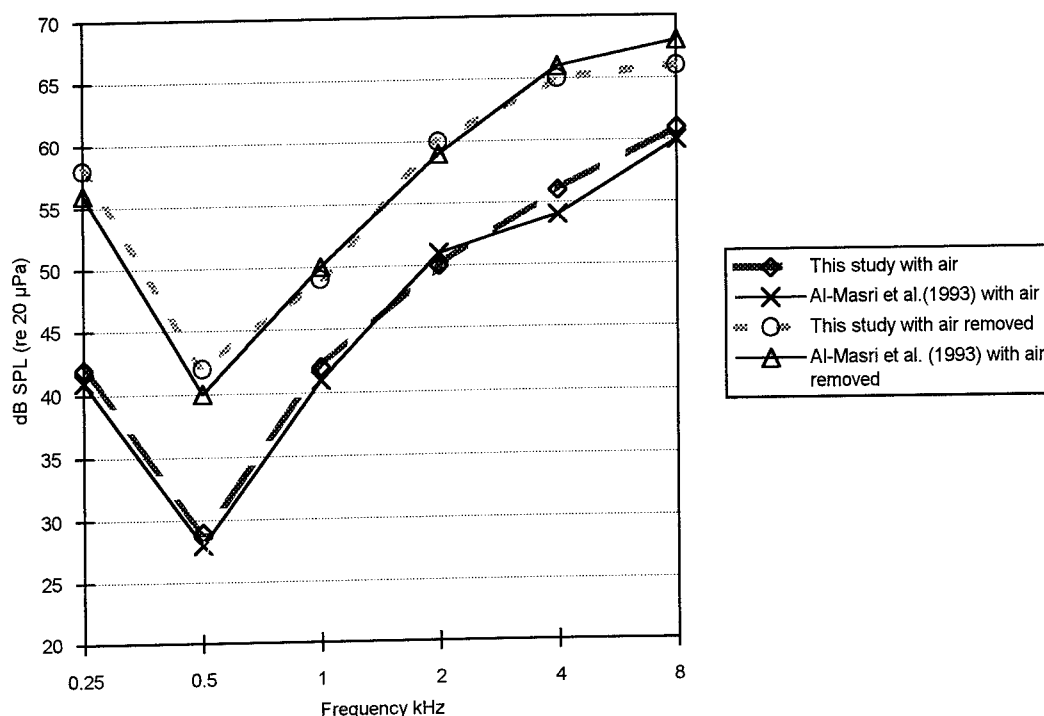


Figure 11-3 A comparison between the mean values of underwater MAF for this study and the pilot study.¹⁰

The maximum ear sensitivity underwater (lowest MAF value) is located around 0.5 kHz. The underwater MAF with air trapped in the ear canals is 42 dB SPL at 0.25 kHz decreasing to a minimum value of 29 dB at 0.5 kHz followed by an increase to a maximum of 61 dB at 4 kHz. The underwater MAF curve with air removed from the ear canals is significantly higher ($p < 0.05$) by 5–17 dB at all frequencies than the MAF curve with air in the ear canals, except at 8 kHz where $p = 0.052$.

Discussion

To evaluate the reliability of the underwater threshold measurement procedure, the mean underwater MAF curves, with air present and with air removed from the ear canals, are presented graphically together with those of a pilot study¹¹ in Figure 11-3. It is clear from these results that there is a good agreement ($p > 0.05$) between the two studies at all frequencies. Hence, it is apparent that the meth-

odology developed¹⁰ for testing underwater MAF provides repeatable results.

Figure 11-4 presents a comparison between the MAF with air in the ear canals for this study and previous studies for bareheaded diving conditions without removing the naturally trapped air from the ear canals. It can be seen from this figure that the results of this study generally support the previous indications that the ear underwater is less sensitive compared with air and that the shape and values of the MAF curve underwater are also different from that in air. This supports our proposed hypothesis that the threshold of hearing underwater is different from that in air. Current noise exposure limits in an air environment are not directly applicable to underwater conditions and suitable noise exposure limits need to be developed. However, Figure 11-4 further reveals that the underwater MAF curve found is quite different in shape and degree from previous studies of underwater hearing. The MAF curve found in this study is dramatically lower than that of the

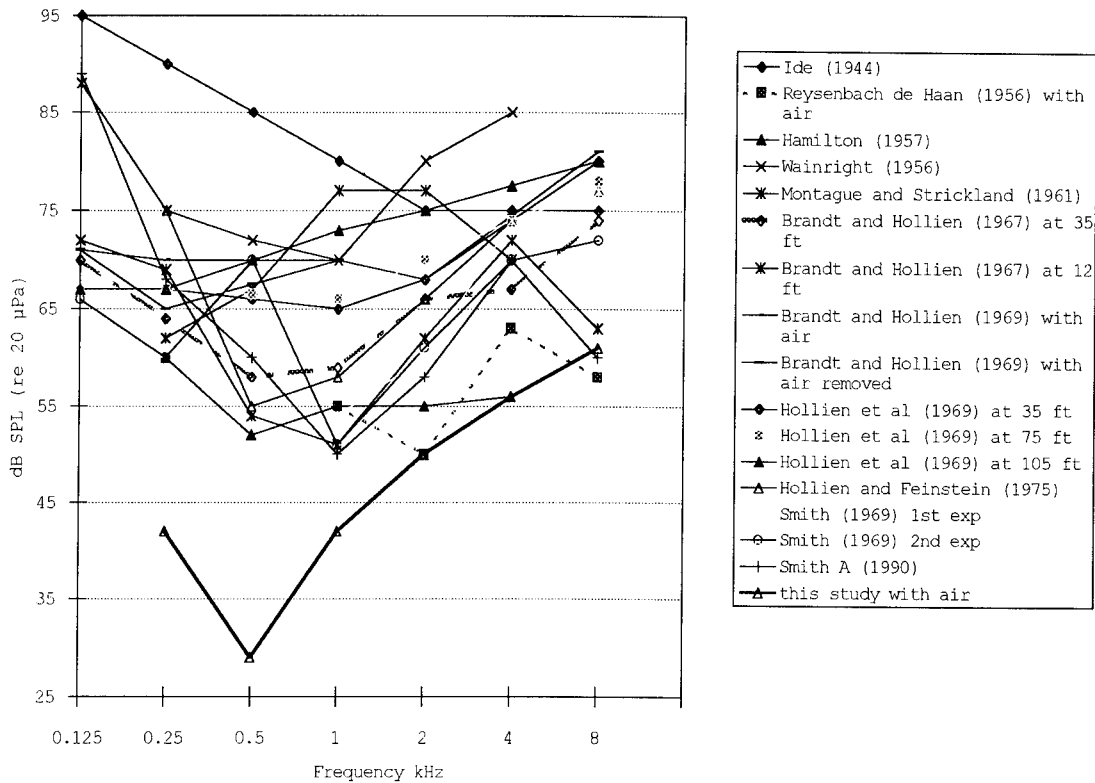


Figure 11-4 Comparison between the MAF with air in the ear canals for this study and previous studies for bareheaded diving conditions.

majority of the previous studies by 18–50 dB at 0.25–0.5 kHz, and by 8–40 dB at 1–8 kHz. This study also shows that the ear is more sensitive at low frequencies than at high frequencies, with the maximum sensitivity being at 0.5 kHz. On the other hand, it is apparent that the MAF curves of the previous studies show a tendency of the ear to be more sensitive at midfrequencies (0.5–2 kHz) with a maximum sensitivity of 1 kHz, rather than at low and high frequencies. This discrepancy is likely to be due to a lack of appreciation of underwater ambient noise by previous researchers that evidently masked the thresholds at low frequencies.

Further, it can be noted from Figure 11-4 that the MAF of this study and the majority of the previous studies increase above 1 kHz with increasing frequency to maximum values at 8 kHz. The MAF values of previous studies are approximately the same (to within 10 dB) in the range 4–8 kHz but 15–20 dB above the curve of this study. The similarities in the

shapes of the MAF curves, particularly at 4 and 8 kHz, indicate that the results of previous studies may more closely reflect the threshold measurements of the present study. Possible reasons for the differences between this study and others are those discussed in the beginning of the chapter.

Additionally, because the MAF curves of Brandt and Hollien¹⁷ and Kirkland et al.²⁰ were adopted by the US Navy as the basis for proposed underwater noise exposure limits,^{20,22,23} a comparison between these and the MAF curve of this study is presented graphically in Figure 11-5. It can be seen from this figure that the MAF curve of this study is considerably lower than those of Brandt and Hollien, and Kirkland et al. by 10–30 dB at all frequencies except at 4 kHz. This means that the noise exposure limits proposed by the US Navy and Kirkland et al. underestimate the risk of damage to hearing due to underwater noise exposure. Therefore it can be concluded

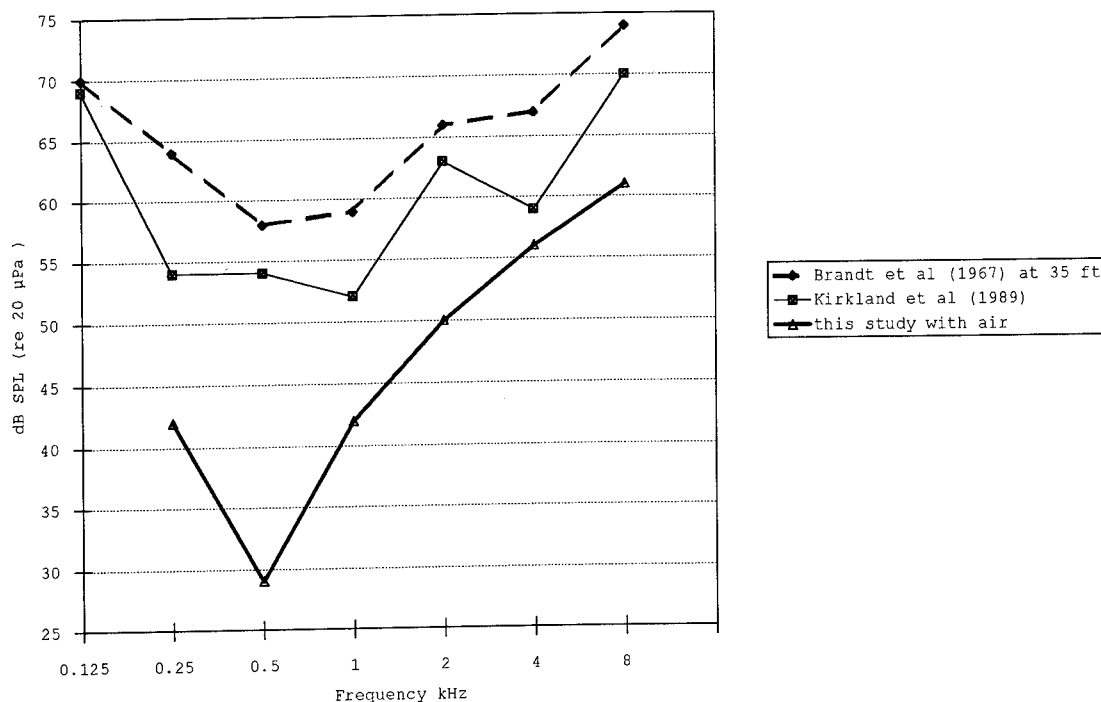


Figure 11-5 Comparison between the MAF with air for this study and that of Brandt and Hollien¹⁷ and Kirkland et al.²⁰

that the proposed limits do not accurately reflect safe noise exposures.

Underwater Hearing Mechanisms

Introduction

Normally, in air, sound is heard via two routes, the air conduction pathway and the bone conduction pathway. Air conduction is the primary route for hearing because the impedance mismatch between air and the tympanic membrane is much smaller than that between air and the bones of the skull (by about 40 dB). However, underwater, the primary hearing route may be different from that in air because the impedance mismatch between water and tympanic membrane is much higher than that between water and the bones of the skull.

Knowledge of the mechanisms of underwater hearing is necessary before setting up underwater noise exposure limits for three main reasons:

1. to explain any change in hearing threshold underwater;
2. to explain the improvement in hearing sensitivity underwater with the presence of air in the ear canals;
3. to assess whether current hearing protectors in air are suitable for underwater use.

Underwater hearing mechanisms have been mainly examined in the past in conjunction with studies that measured thresholds. Three pathways have been proposed to explain how sound is transmitted from water to the cochlea. These are the auricular conduction pathway,²⁴⁻²⁷ the bone conduction pathway,²⁸⁻³² and the dual conduction pathway.^{5,14-16,33} However, it is apparent from a review of the literature that the experimental validity of these studies is open to question mainly for two reasons:

1. the methodology and materials used were not suitable to evaluate the underwater hearing mechanisms;

2. the majority of the studies failed to report important experimental details, such as the underwater ambient noise levels and the bone conduction threshold of the subjects.

Therefore, in order to study this problem it is important to develop methods capable of evaluating the roles of bone conduction and auricular conduction pathways separately. An assessment of the importance of auricular conduction would require acoustical isolation of the bones of the skull from water borne sound transmission. Ideally, this could be achieved by use of a material that attenuates sound transmission from water to bone without affecting sound transmission through the external ear canals. Unfortunately, because the external auricular canals are embedded in the temporal bone, this approach is in practice impossible to achieve.

On the other hand, an evaluation of the importance of the bone conduction pathway would require attenuation of the sound transmission from water through the auricular conduction pathway to the cochlea. This can be achieved through two different approaches. The first is by comparing the underwater hearing thresholds of subjects with conductive hearing loss and normal hearing. This is because the reduction in sound pressure transmission to the cochlea through the auricular conduction pathway is more likely to remain constant in air and water. Hence, if the bone conduction pathway is important for hearing underwater the difference in hearing thresholds in air between the subjects with normal hearing and those with conductive hearing loss would be expected to disappear underwater. Conversely, if the bone conduction pathway does not participate in underwater hearing, the difference in thresholds between the normal hearing subjects and those with conductive hearing loss will remain constant in both media. Unfortunately, it is likely to be extremely difficult to find practicing divers with conductive hearing loss, due to diving safety regulations that prevent diving with any significant hearing loss. The second approach involves blocking the ear canals with a

material that attenuates sound transmission from water through the auricular pathway. It is important that this material should not affect the incident sound field on the skull, otherwise the results will be meaningless. However, the practical difficulties of finding a suitable material to block the ear canals and meet the criteria discussed above is not a simple task. Therefore, both approaches were adopted, and an intensive search for a suitable material and for divers with conductive hearing loss was conducted.

Underwater Hearing Thresholds With Ear Plugs

Methodology

It became evident after a thorough theoretical and experimental search that steel ear plugs were practical to use underwater for blocking the ear canals.¹⁰ The experiment was carried out using 24 normally hearing sport divers (18 males and 6 females). The subjects were selected after screening audiometry and impedance, to ensure that they were otologically normal. All the subjects were inexperienced in hearing threshold measurements. Thresholds were measured in air and underwater with and without the ear plugs using $1/3$ -octave band noise stimuli and sound field audiometry. A headband made of plastic (characteristic acoustic impedance is similar to water³⁴) was used to keep the plugs in place, both in air and underwater. When necessary several ear plugs were tried in order to find the best acoustic fit. This was judged subjectively. Each subject then kept the same ear plugs throughout testing. For underwater measurements with plugs, the subjects fitted them in air before submerging underwater. The procedures and equipment are those described previously.

Results and Discussion

The mean MAF results in air with and without the ear plugs and the difference between them are plotted in Figure 11-6. This figure shows that in air the hearing thresholds of the sub-

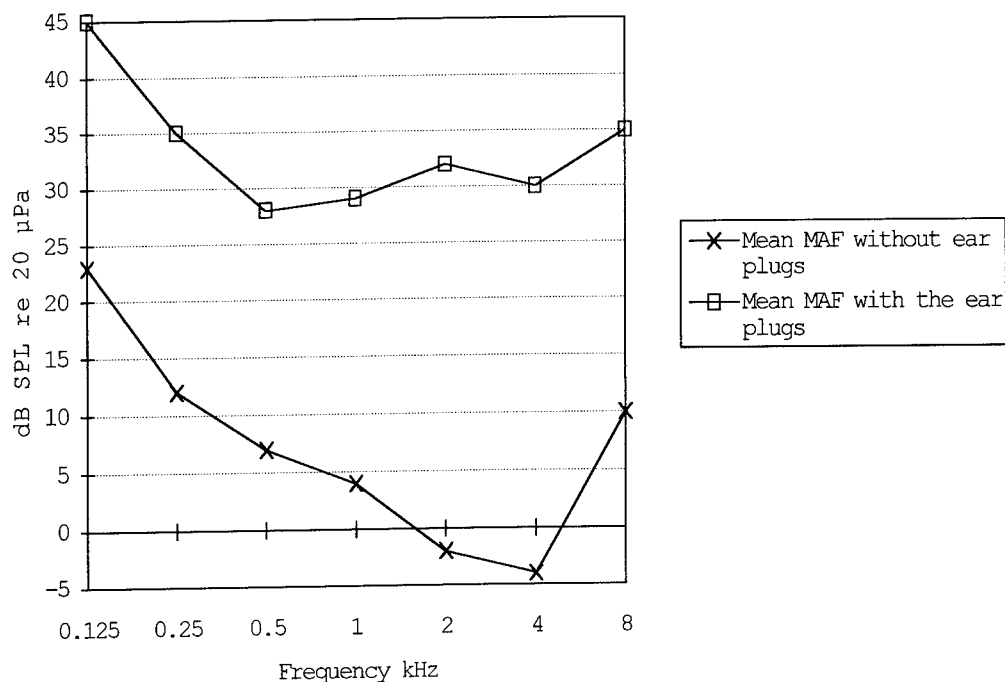


Figure 11-6 The mean difference in air between the MAF using $1/3$ -octave band noise with and without steel ear plugs.

jects are dramatically increased ($p = 0.000$) when the subjects wore the steel ear plugs. The mean difference between the thresholds with and without the plugs is 21 dB (range 10–35 dB) at 0.0125–0.5 kHz and increases with increasing frequency to a maximum of 35 dB (range 25–45 dB) at 4 kHz, followed by a decrease to 25 dB (range 9–40 dB) at 8 kHz. These results agree with the theoretical prediction.¹⁰ This demonstrates that the steel ear plugs are effective and fit all the subjects' ears. Consequently, it was expected that for subjects wearing the same ear plugs underwater, their thresholds would increase by at least the same value as in air if the auricular conduction pathway is the only route for sound transmission from water to the cochlea.

Figures 11-7 and 11-8 show the mean underwater thresholds with and without the plugs, as well as the mean differences between them. It is clear that the steel ear plugs have a negligible effect on the underwater MAF. The curves with and without the plugs are the same ($p > 0.05$), to within 4 dB, at all frequencies. The results indicate that attenuation of the sound

pressure transmission through the auricular pathway does not have any effect on the underwater MAF. This is strong evidence supporting the hypothesis that the bone conduction pathway is important in underwater hearing mechanisms.

Figure 11-9 presents a comparison between the underwater MAF with air in the ear canals (without the steel ear plugs) and the MAF of the previous study. It is clear from this figure that the results are remarkably repeatable ($p > 0.05$), to within 3 dB, at all frequencies. This emphasizes the conclusion drawn previously that the facilities and methodology developed for underwater MAF measurements give repeatable and reliable results.

Underwater MAF With Conductive Hearing Loss

As a result of an intensive search for divers with a significant conductive hearing loss, three subjects were found. Two of them had bilateral moderate conductive hearing loss, probably due to bilateral otosclerosis, and one

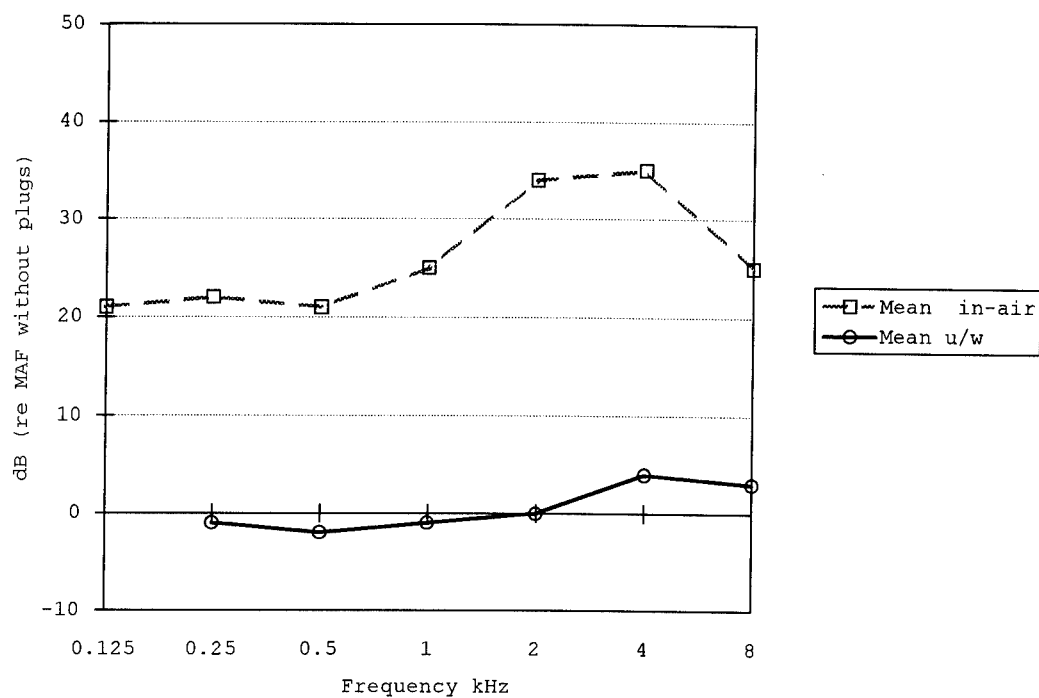


Figure 11-7 The mean difference in air and underwater between MAF with and without the steel ear plugs.

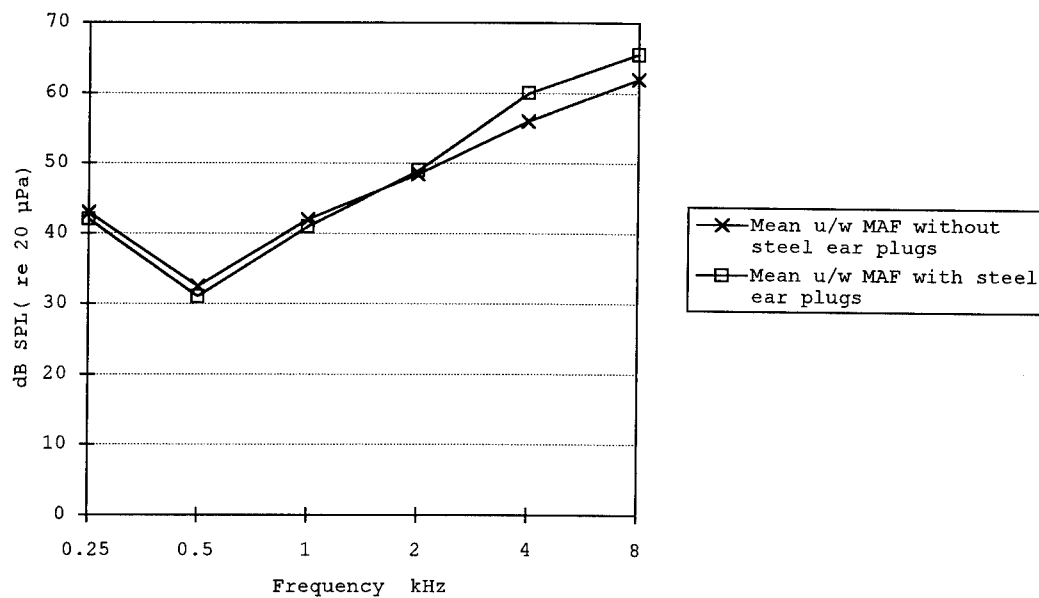


Figure 11-8 Underwater MAF curves with and without the steel ear plugs.

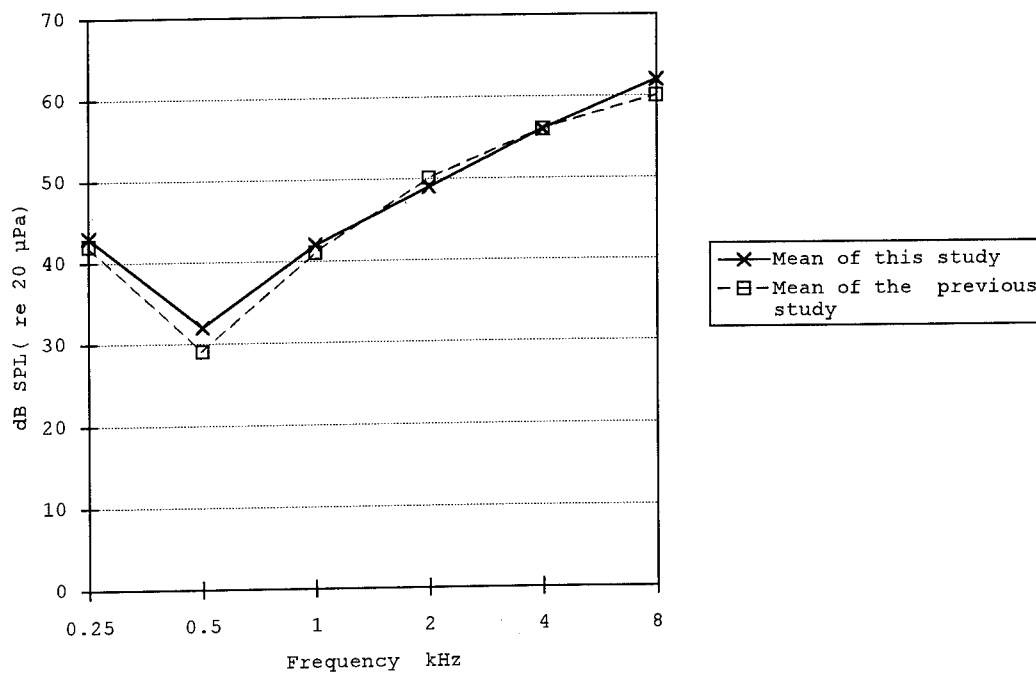


Figure 11-9 Comparison between the underwater MAF with air in the ear canals (without the steel ear plugs) and the MAF of the previous study presented in Figure 11-4.

subject had unilateral moderate conductive hearing loss of unknown cause. These subjects were used to reexamine the importance of the bone conduction pathway in underwater hearing mechanisms. Because it is well known that in air the occlusion effect and the participation of ossicular inertia in bone conduction hearing thresholds disappear with otosclerosis, it was considered a unique opportunity to test the underwater MAF of these subjects with and without air removed from the ear canals. This was done in order to test the hypothesis that the improvement in underwater MAF with the presence of air in the ear canals is due to the "occlusion effect" and the participation of the "ossicular inertia" in underwater hearing thresholds. If these are the reasons for the improvement in underwater MAF with air in the ear canals, this improvement is expected to disappear with conductive hearing loss.

The subjects' hearing thresholds and middle ear function in air were thoroughly investigated using pure-tone audiometry (air and bone conduction, and masking when needed)

and impedance measurements (tympanometry, stapedial reflex, and Eustachian tube function). Their MAF was measured in air and then underwater. The underwater MAF of each of the two subjects with bilateral conductive hearing loss was tested with air present and with air removed from the ear canals. The subject with unilateral conductive hearing loss was tested with air present and with air removed from both ear canals, with air removed from the left ear only, and with air removed from the right ear only. The subjects had a 10 minute break at midtime to maintain their concentration.

Results and Discussion

The air conduction hearing threshold curves of the two subjects who had bilateral otosclerosis were 30–50 dB hearing loss (HL) at 0.25 kHz and these decreased with increasing frequency to a minimum of 25–30 dB at 2 kHz. Then the curves increased with increasing frequency to a maximum of 40–50 dB at 8 kHz. The bone conduction curves at –10 dB HL to

5 dB for all frequencies except at 2 kHz where the thresholds are 15–20 dB. The gaps between the air conduction curves and bone conduction curves at all frequencies are 20–50 dB except at 2 kHz where the gap is 10 dB. The bone conduction hearing loss at 2 kHz is most likely due to the otosclerosis phenomenon known as the “Carhart notch.” This is due to a mechanical effect whereby the ossicular inertia is lost at 2 kHz and does not participate in hearing through the bone conduction pathway. The results of the impedance measurements were also consistent with otosclerosis.

The results of pure-tone audiometry and impedance measurements on the third subject (with a unilateral conductive hearing loss) show that the right ear had normal hearing thresholds, whereas the air and bone conduction curves were almost equal to 0 dB HL at all frequencies. The left ear had a conductive hearing loss. The air conduction curve was flat 25–30 dB HL at all frequencies, except at 4 kHz where it rose to 10 dB, and the bone conduction curve was also flat and almost 0 dB HL. The middle ear pressure of both ears was normal and consistent with a patent Eustachian tube. The compliance of both ears was within normal limits but with a slight asymmetry, with the right ear compliance (0.7 mL) being lower than the left ear (1.2 mL). The acoustic reflexes of the right ear were present but elevated (100–110 dB HL) for contralateral stimulation (probe right ear and signal left ear). The acoustic reflexes of the left ear were absent.

The results of the two subjects with bilateral conductive hearing loss show that the conductive hearing loss disappears underwater. This is strong evidence supporting the hypothesis that the bone conduction pathway is important in underwater hearing. A more important finding, perhaps, is that the bone conduction hearing loss due to the Carhart notch in air remains underwater. This is evidence that the ossicular inertia component participates in underwater hearing the same as in bone conduction hearing in air. The underwater MAF curves with air and with air removed are the same (within 5 dB) at all frequencies, that is, the influence of removal of air from the ear canals on the MAF disappeared. This sup-

ports the hypothesis that the improvement in underwater MAF with the presence of air in the ear canals is mainly due to the occlusion effect and ossicular inertia. This also means that the external ear canals participate in underwater hearing in the same way as bone conduction hearing mechanisms in air.

It can also be noted from inspection of Figure 11-10, which shows the underwater results of the subject with unilateral hearing loss, that the MAF curves, with air present and with air removed from both ear canals, are equal (to within 3 dB) to the respective reference MAF presented in Figure 11-4. When air is removed from the right ear (normal ear) only, that is, without air removed from the left ear (with conductive hearing loss), the underwater MAF curve is equal (within 3 dB) to the reference MAF curve when air is removed. On the other hand, when the air is removed from the left ear (conductive hearing loss) only, the underwater MAF curve remains equal (within 3 dB) to the reference MAF with air. These results indicate that the effect of removing air from ear canals disappears with conductive hearing loss and that the external ear canals participate in underwater hearing mechanisms. This is further evidence supporting the hypothesis that the presence of air in the ear canals improves underwater hearing thresholds due to the occlusion effect and ossicular inertia.

Thus, it can be concluded that hearing underwater, similar to bone conduction hearing, occurs via contributions of the external ear canal, the middle ear through the ossicular inertia, and the inner ear. This conclusion is important and helps clarify our understanding of the mechanisms of hearing underwater through the bone conduction pathway. Previously the external ear and the middle ear were not thought to contribute to underwater hearing.^{9,10}

Another important implication for these results, which show that bone conduction is important in underwater hearing, is that conventional hearing protectors in air are likely to be ineffective if used underwater. Thus, new protectors for underwater use need to be considered.

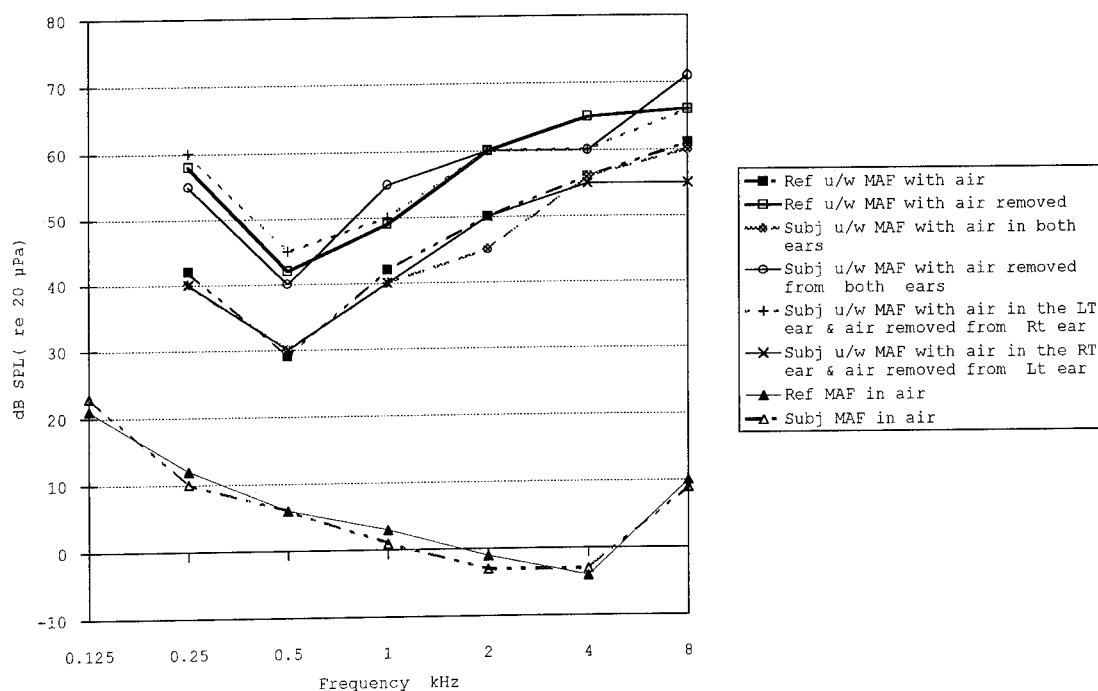


Figure 11-10 Comparison between the MAF in air and underwater, with air and with air removed from the ear canals of subject no. 3 with unilateral conductive hearing loss (CHL) and that of the MAF curves shown in Figure 11-4.

Underwater Noise Exposure Limit

It is evident from the results of the underwater hearing thresholds and hearing mechanisms that the current noise exposure limits, hearing damage risk criteria, and the A-weighting scale applicable in air are not suitable for underwater use. Therefore noise limits and hearing damage risk criteria need to be developed for underwater use^{35,36}. It was discussed in the introduction that these may be achievable by modifying the A-weighting scale to produce an equivalent weighting scale for underwater use. This would involve subtraction, at each frequency, of the difference between the MAF values for $1/3$ -octave band noise in air and underwater (established in Figure 11-4) from the A-weighting scale. This assumes the relationship between the 40 phon curve and the MAF curve at each frequency is constant in air and in water. The new underwater scale will be called the W-weighting scale. Table

11-1 illustrates the procedures used to derive the W-weighting scale. Thus the current accepted industrial noise limits can be applied to the underwater situation in terms of the W weighting.

The inverse of the A-weighting scale and W-weighting scale and the MAF curves in air and underwater are plotted in Figure 11-11. It can be seen from this figure that the A-weighting scale accounts for the increased sensitivity of the ear in air over the frequency range 1–6 kHz. In contrast, the W-weighting scale accounts for the increased sensitivity of the ear underwater over the frequency range 0.25–1 kHz.

It is apparent from published work that the explanation of the cause for noise-induced hearing loss in air at 4 kHz applies to noise-induced hearing loss due to underwater noise exposure that occurs at the midfrequencies around 1 kHz; this is about half to one octave above 0.5 kHz. The maximum sensitivity of

Table 11-1 Mathematical Steps Used to Derive W-Weighting Scale From A-Weighting Scale

Frequency (kHz)	0.25	0.5	1	2	4	8
MAF underwater (dB SPL)	42	29	42	50	56	61
MAF of 1/3-octave band noise in air	12	6	3	-1	-4	10
Δ (dB) = MAF underwater - MAF in air	30	23	39	51	60	51
A-weighting scale (dB SPL)	-8	-3	0	1	1	-1
W-weighting scale = A - weighting - Δ	-38	-26	-39	-50	-59	-52

Data are rounded to the nearest 1 dB.

the ear underwater is located at the low frequencies around 0.5 kHz, where the hearing threshold is 20–30 dB less than at 2–4 kHz. Because it is well known that the midfrequencies are more important for understanding speech than frequencies around 4 kHz, noise-induced hearing loss due to underwater noise exposure may result in greater hearing disability than that due to noise exposure in air. This

has two important implications. First, the current hearing damage risk criteria in air may be unsuitable for underwater use. Second, direct transposition of the current noise limits from air to underwater may provide less protection against hearing disablement than the current limits for air. Therefore, considerable caution needs to be exercised in applying limits expressed in terms of dB(W).

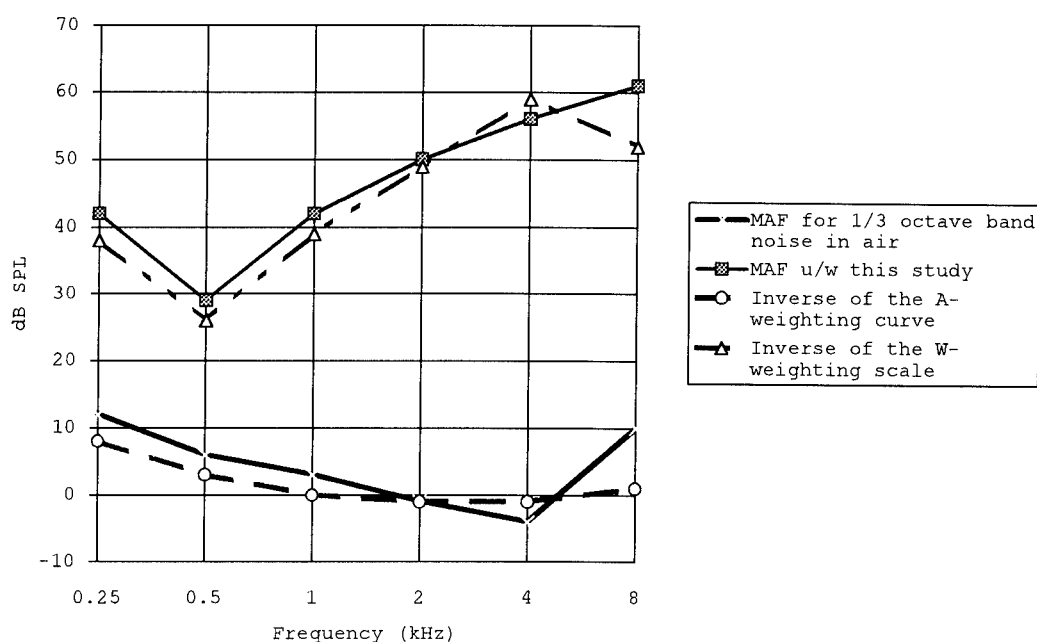


Figure 11-11 Relationship between MAF in air, MAF underwater, the inverse of the A-weighting curve, and the inverse of the W-weighting curve.

Conclusions

Underwater MAF curves for $1/3$ -octave band noise, with air and with air removed from the ear canals, have been established over the frequency range 0.25–8 kHz. It was found, similar to the previous studies, that the ear underwater is less sensitive than that in air, but the results of this study are 20–35 dB lower. The removal of the trapped air from the ear canals increase the underwater MAF by 7–15 dB. The underwater MAF curves (with air and with air removed) of this study are dramatically different in shape and value from that in air. The maximum sensitivity of the ear underwater is located around 0.5 kHz. Possible reasons for the difference between MAF values in air and underwater include water loading mass on the ear drums, as well as diminished head diffraction gain, external ear resonance, and middle ear amplification when underwater.

Underwater hearing mechanisms were investigated using steel ear plugs and subjects with conductive hearing losses. It was shown that the bone conduction pathway is the primary route for underwater hearing. The external ear canals and middle ear ossicles participate in underwater hearing in the same way as they participate in the bone conduction pathway in air.

A W-weighting scale, equivalent to the A-weighting scale in air, was developed to assess the risk of damage to hearing from underwater noise. It was proposed that the current industrial noise limits can be applied to the underwater situation in terms of the W-weighting scale. However, until further research is carried out, considerable caution still needs to be exercised.

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Chapter 12

Threshold Shift Dynamics Following Interrupted Impact or Continuous Noise Exposure: A Review

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and Kelly A. Underwood

Over the last few years a number of experiments have confirmed the observations of Miller et al.¹ that threshold shift (TS) in subjects given a brief daily noise exposure over several days could recover as much as 30 dB despite the continuing exposure cycle. This effect is now commonly referred to as a toughening effect. Another manifestation of toughening was demonstrated by Canlon et al.² who showed that an exposure to a low level conditioning noise could reduce the permanent threshold shift (PTS) from a subsequent high level exposure despite the absence of an effect from the conditioning exposure. While the experimental paradigms for these two experiments are quite different, the protective effects that they produce are thought to involve similar cochlear mechanisms. Interest in interrupted noise exposure paradigms was revived after Clark et al.³ confirmed the Miller et al.¹ results. A small body of literature has emerged since then showing that interrupted exposures using broadband or narrowband, continuous or impulsive noise could elicit a cochlear toughening effect. These studies have involved the use of behavioral conditioning, brain stem evoked potentials, gross cochlear potentials, single VIII nerve recordings, and distortion product emissions to document the phenomena. This chapter reviews the existing body of data acquired from interrupted noise exposure paradigms.

Review of Results from Interrupted Noise Exposure Paradigms: Toughening Effects

Miller et al.¹ used a 115 dB sound pressure level (SPL) broadband of noise presented on a 7.5 minute daily cycle for 16 days to cats that were behaviorally trained in order to acquire an estimate of pure-tone thresholds. A summary of their threshold data obtained at the 4.0 kHz test frequency is shown in Figure 12-1. In this figure the filled square refers to the mean threshold shift of about 45 dB measured following a single 7.5 minute exposure. After a month of recovery, thresholds returned to normal and the group was then exposed to the same noise for 16 consecutive days (solid circles). After the second day, thresholds had shifted about 40 dB, similar to the shift measured a month earlier. However, by the fifth day of the exposure, thresholds had recovered about 30 dB and remained relatively stable until the termination of the 16 day exposure cycle. After a 4 month recovery, during which time thresholds returned to near normal, the animals were again exposed to the same noise cycle, this time for only 8 days. The exact TS values were not presented, but Miller et al.¹ indicated that they were about the same as those measured on the last 7 days of the 16 day exposure cycle (these are indicated by the flags on the symbols in Figure 12-1a. Thus, the

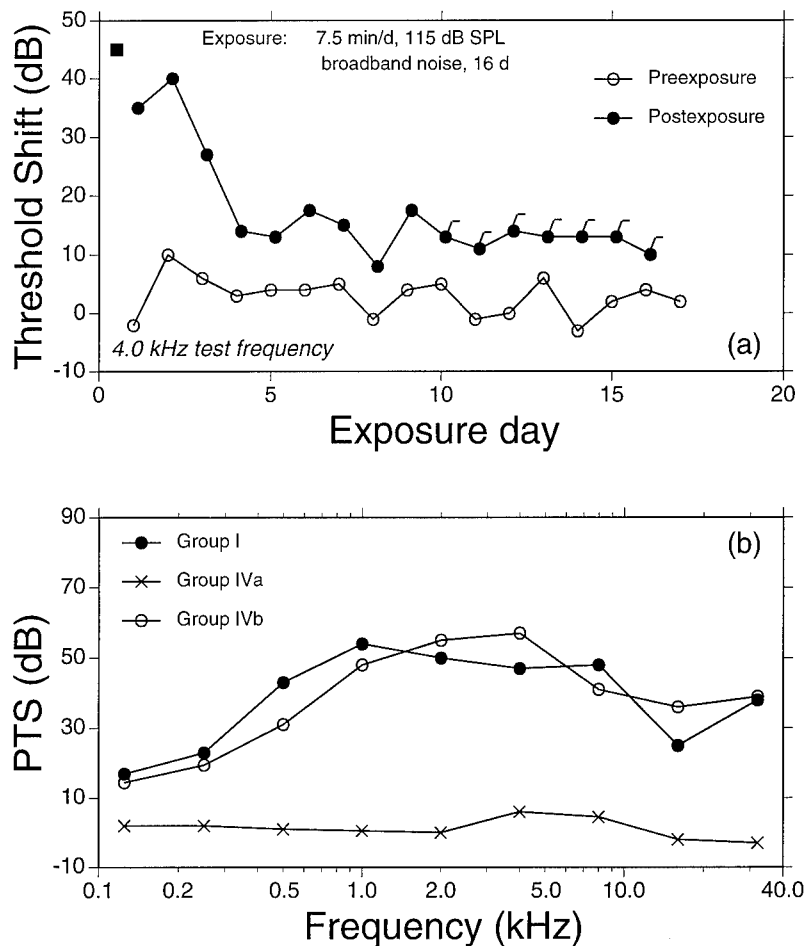


Figure 12-1 The results of the Miller et al.¹ study of interrupted continuous noise exposures. **(a)** The recovery of threshold at the 4.0 kHz test frequency despite the continuing exposure. **(b)** The permanent threshold shift (PTS) audiograms that show that the "toughened" animals were equally susceptible to noise trauma as were "untoughened" animals (closed versus open circles).

30 dB toughening effect evidently lasted for at least 4 months. There was little PTS (~5 dB at 4.0 and 8.0 kHz; see Figure 12-1b, × symbol) following these interrupted exposures. After another month, animals from this group were exposed to the same noise but for two uninterrupted hours. The animals whose PTS audiogram is shown in Figure 12-1b (open circles) showed the same PTS as did a group exposed to the same noise but without a history of prior interrupted exposures (solid circles). The toughening effect elicited by the interrupted exposure did not seem to offer any protection to the subjects exposed to the 2 hour expo-

sure. This result is not in accord with the nature of the toughening effect produced by the low-level noise-conditioning paradigm of Canlon et al.²

In the Canlon et al.² experimental paradigm guinea pigs were exposed to a 1.0 kHz, 81 dB SPL pure tone for 24 days (conditioning exposure). This exposure produced little or no TS or PTS as measured by evoked potentials recorded from subcutaneous scalp electrodes. The conditioning exposure was followed by a 1.0 kHz, 105 dB SPL tone for 72 hours. A reference group received only the 105 dB SPL exposure. A summary of their results is

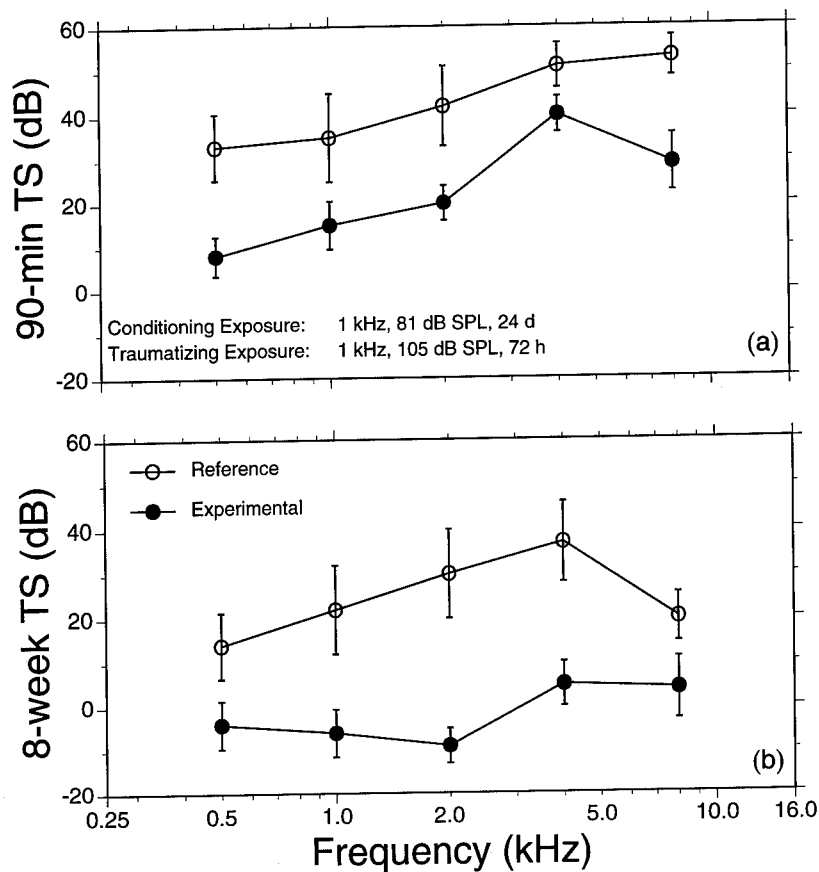


Figure 12-2 (a) The mean threshold shift (TS) measured nine minutes following exposure to a 1.0 kHz tone at 105 dB SPL for 72 hours in groups of animals that were either conditioned (●) or not conditioned (○) by a previous low level noise exposure. (b) The mean threshold shifts from the same two groups of animals following an eight-week recovery period. The conditioned group (experimental) showed substantially less hearing loss across the frequency range tested. From Canlon et al.²

shown in Figure 12-2. Within 90 minutes of the termination of the 105 dB exposure, the conditioned group showed up to 30 dB less TS than the unconditioned group and after 2 months these differences, which probably reflect differences in PTS, were even greater. Thus, the conditioning exposure apparently toughened the system and the toughening is reflected in TS measured during the early postexposure period when TSs are rapidly changing as well as in PTS after TSs have stabilized.

At least part of the effects described above were thought to be the result of the acoustic

reflex mechanism. However, Ryan et al.⁴ and Henderson et al.⁵ have recently shown that this is not the case and that the toughening effect is probably the result of cochlear mechanisms.

Saunders et al.⁶ also reported on the results of an interrupted noise exposure paradigm. They used an octave band of noise (OBN) centered at 4.0 kHz at levels that varied from 57 to 92 dB SPL. The exposures lasted 6 h/d for 10 consecutive days. Their threshold data for the 86 dB SPL exposure is shown plotted in Figure 12-3a for the 5.7 kHz test frequency. Both the daily postexposure TS (solid circles)

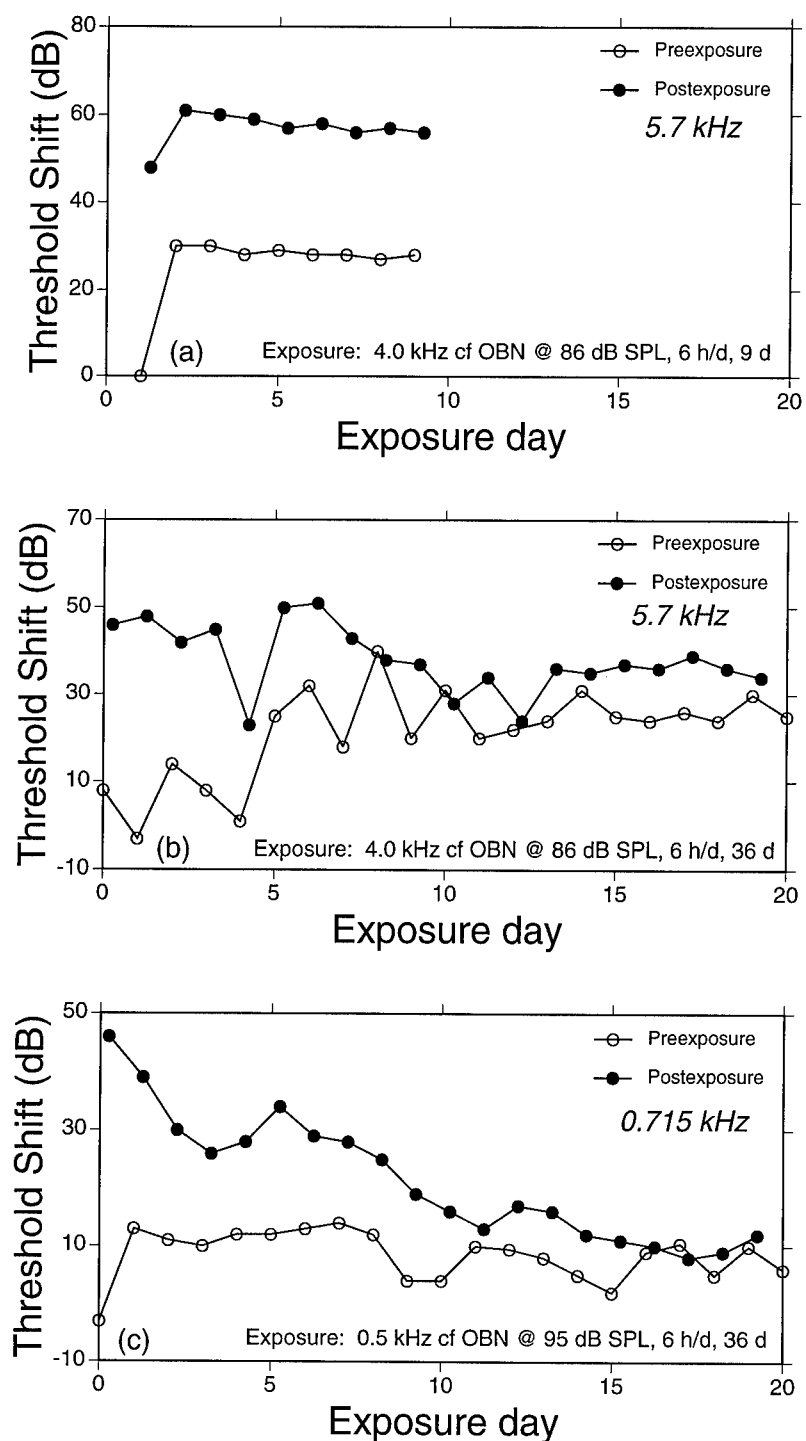


Figure 12-3 The daily group mean threshold shift at the indicated test frequencies following various interrupted noise exposures in the chinchilla. (a) From Saunders et al.⁶ (b) From Clarke and Böhne.⁷ (c) From Clark et al.³

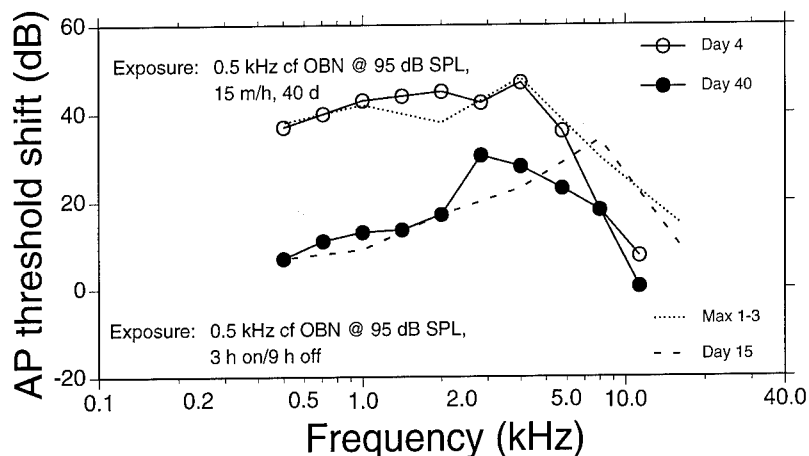


Figure 12-4 Threshold shifts of the whole nerve action potential (AP) following interrupted noise exposures. In these two studies, the same noise was used but on a different presentation schedule. In both data sets the “toughening” effect is clearly evident at the lower frequencies. From Sinex et al.⁸ and Boettcher et al.⁹

and the preexposure TS (open circles) were quite stable over the 10 day exposure cycle. Similar results were found for the other exposure levels. Using a similar exposure paradigm and noise stimulus, except on a 36 day cycle, Clark and Bohne⁷ showed that after about the first 6 days when TS was relatively stable, TS began to decrease slightly (see Figure 12-3b) so that by day 10 thresholds had recovered about 10 dB and TS then remained stable through day 36 of the exposure cycle. However, when Clark et al.³ used an OBN centered at 0.5 kHz at 95 dB SPL in the same exposure paradigm, a large (>30 dB) and consistent decrease (Figure 12-3c) in TS was observed over the first 10 days of exposure at a half-octave above the stimulating noise.

The unusual dynamics of these behaviorally determined TSs were confirmed by Sinex et al.⁸ and Boettcher et al.⁹ using electrophysiological techniques. The former showed that the toughening effect was manifested in the whole nerve action potential (AP) as well as in single VIII nerve recordings and in tuning curve functions. An example of results from Sinex et al.⁸ is shown in Figure 12-4. Chinchillas were used as the experimental subject and a 0.5 kHz cf OBN at 95 dB SPL presented

on a 15 min/h schedule over 40 days was used as the exposure stimulus. The AP recorded on the fourth day of exposure is contrasted with that recorded on day 40 in figure 12-4. A clear reduction in TS by day 40 of up to 30 dB across the 0.5 through 2.0 kHz region was measured; about the same degree of toughening was obtained with behavioral thresholds. In the Boettcher et al.⁹ experiments, the same noise stimulus was used except on a 3 hours on, 9 hours off cycle over 15 days. Their mean AP recordings from the exposed chinchillas over the 15 day cycle is shown in Figure 12-5a. Again there is a 20 dB or greater toughening effect across the 0.5–2.0 kHz region. When the Boettcher et al.⁹ AP threshold shift audiograms on days 1–3 and those obtained on day 15 are compared with Sinex et al.⁸ the nearly identical functions shown in Figure 12-4 were obtained. Despite the recovery of a threshold over the 15 day cycle of exposures, outer sensory cell loss as shown in Figure 12-5b, as well as a deterioration in the condition of the cilia, was found to systematically increase over the exposure cycle.

The effect of noise level and frequency on the extent of toughening was presented by Subramaniam et al.^{10,11} using the evoked

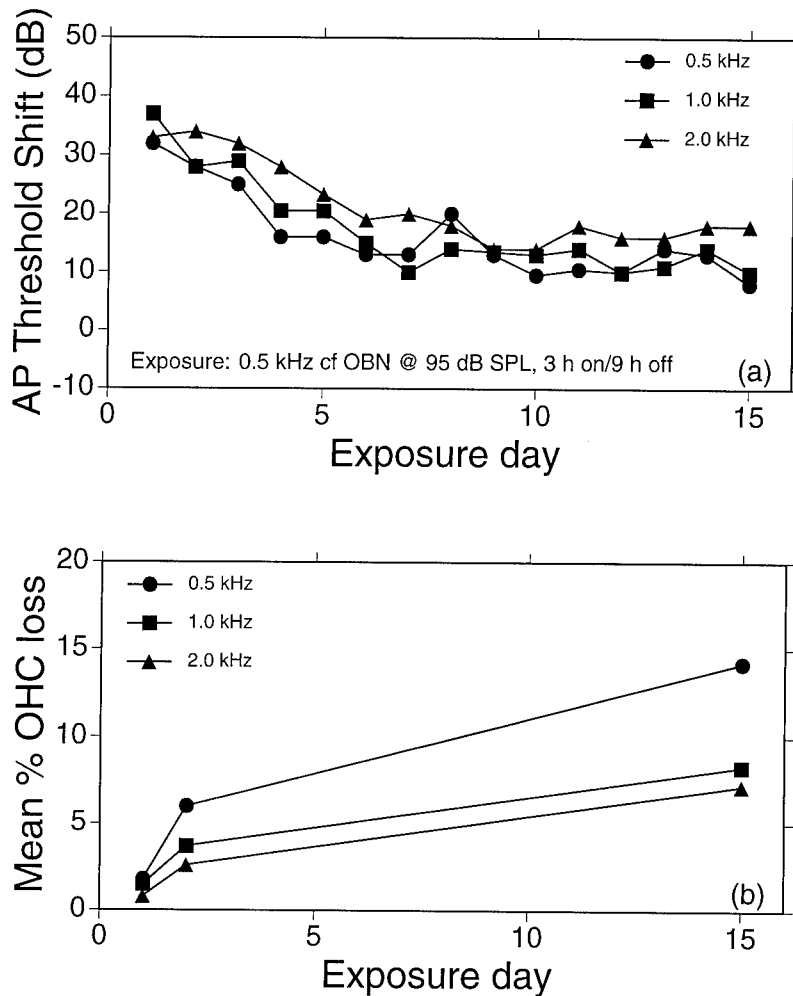


Figure 12-5 (a) The recovery of the whole nerve action potential (AP) threshold during a 15-day interrupted noise exposure at the three test frequencies indicated. From Boettcher et al.⁹ (b) The mean number of outer hair cell (OHC) missing in a 1 mm segment of the organ of Corti at the indicated frequency region following 1, 3, and 15 days of the interrupted exposure.

brain stem potentials in a chinchilla model. A summary of their data is shown in Figures 12-6a and b. The exposure followed a 10 day cycle of 6 h/d: 0.5 kHz cf OBN at 85, 95, or 100 dB SPL or 85 dB SPL, 4.0 kHz cf OBN noises were used as stimuli. A 20–30 dB toughening effect that showed a systematic frequency specificity was seen for both frequencies of stimulation. Figure 12-6 shows the difference between the TS measured following the first day of exposure and the mean TS on days 9 and 10 for each of the four exposure

conditions. For the low frequency, 85 dB SPL stimulation, a 10 dB effect was seen only at 1.0 kHz. As the intensity of the noise increased, the magnitude of the toughening effect increased and spread to adjacent frequencies such that, at the highest level of stimulation, 100 dB SPL, more than 25 dB TS recovery was seen in the 4.0 kHz region, along with TS recoveries of 10–20 dB at the other test frequencies except 16.0 kHz. Subramaniam et al.,¹² using the same experimental conditions as Boettcher et al.,⁹ showed that the recovery

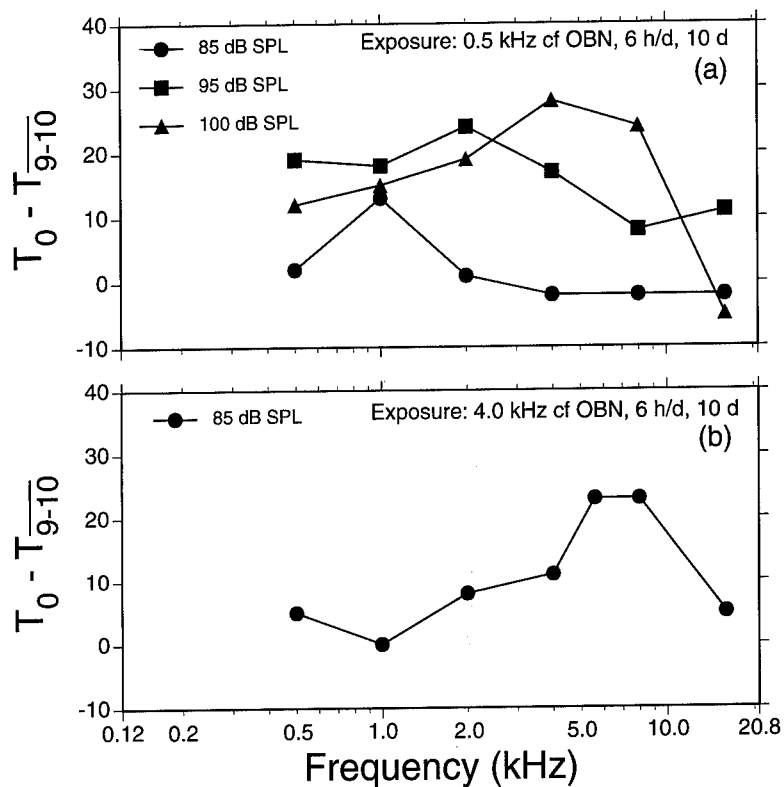


Figure 12-6 The differences between the mean threshold shift measured following the first day of exposure (T_0) and that measured on days 9 and 10 for: (a) three different intensities of an interrupted 0.5 kHz octave band noise (OBN), and (b) an interrupted 4.0 kHz OBN presented at 85 dB SPL. From Subramaniam et al.^{10,11}

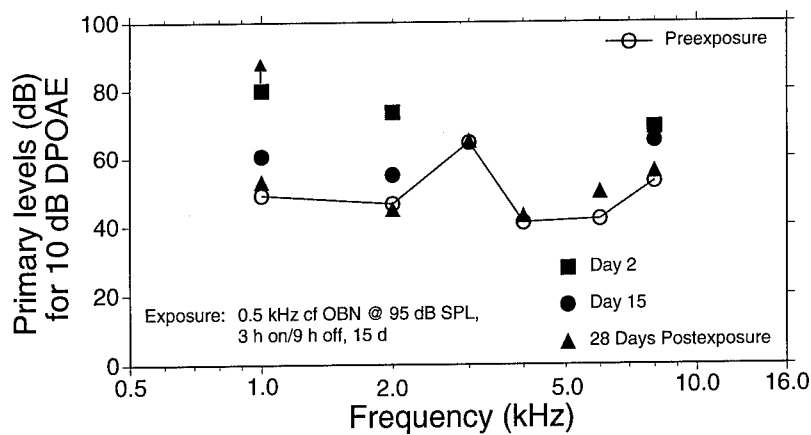


Figure 12-7 Level of the primary tone needed to elicit a 10 dB cubic distortion product ($2f_1 - f_2$) otoacoustic emission at the indicated frequencies $\sqrt{f_1 f_2}$ at various times during the interrupted 0.5 kHz octave band of noise (OBN) (level of f_1 = level of f_2). From Subramaniam et al.¹²

phenomena was also reflected in the $(2f_1 - f_2)$ cubic distortion product otoacoustic emissions (3DPEs). A summary of their 3DPE data is shown in Figure 12-7 replotted as a 10 dB iso-DPE versus geometric mean primary frequency function. The mean preexposure function and 28 day postexposure function are similar indicating recovery of the 3DPE. On the second day of exposure 3DPEs at 1.0, 2.0, and 8.0 kHz were substantially elevated and by day 15 they were close to preexposure

values at 1.0 and 2.0 kHz. At 8.0 kHz there was little change between exposure days 2 and 15.

Interrupted Impact Noise Exposures

Unlike the Miller et al.¹ results, when Henderson et al.¹³ exposed chinchillas to an interrupted broadband, impact noise presented at 113 dB peak SPL on an 8 h/d schedule for 5 days, they showed that the daily TS would

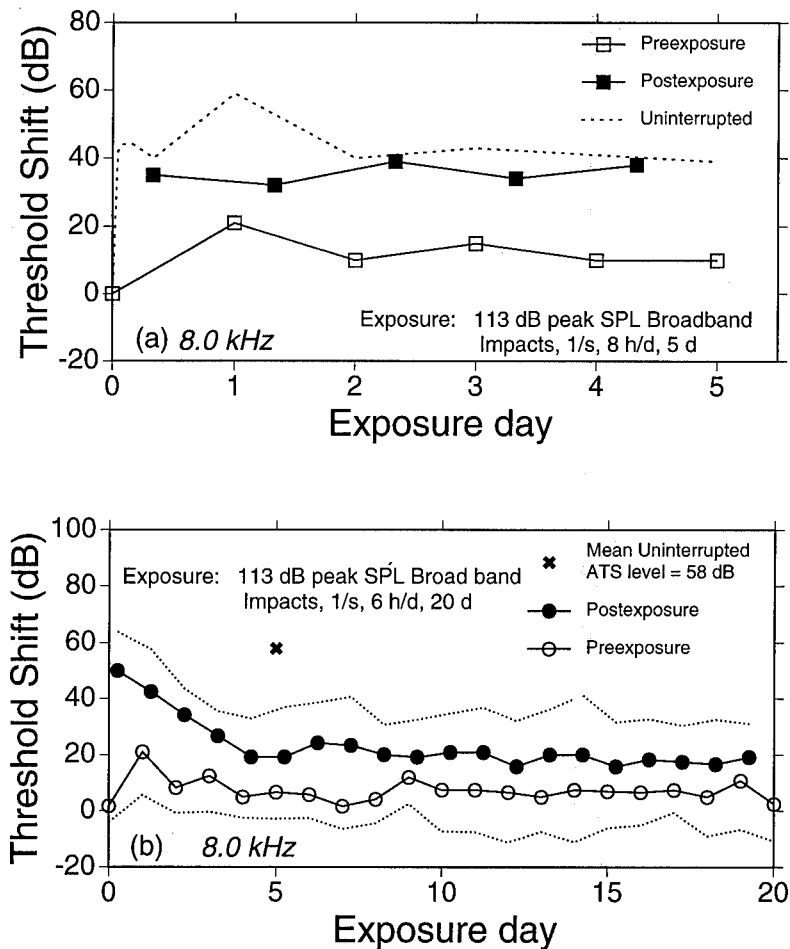


Figure 12-8 (a) The group mean threshold shifts measured at 8.0 kHz prior to and following daily exposure to 113 dB peak SPL impacts on an interrupted schedule, 8h/d for 5d, compared with the threshold shifts following an uninterrupted 5-day exposure. From Henderson et al.¹³ (b) Group mean threshold shifts measured at 8.0 kHz prior to and following exposure to the same impacts but on a 6h/d schedule for 15 days. ✕ refers to the mean ATS level measured in a group of chinchillas exposed for five days to the same impact without interruption. From Hamernik et al.¹⁴

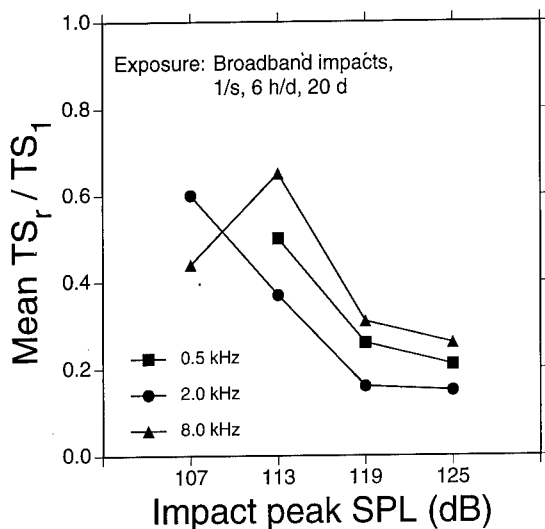


Figure 12-9 The group mean threshold shift recovery fraction, TS_r/TS_1 at the indicated test frequencies for interrupted impact noise exposures having peak levels of 107 through 125 dB SPL. From Hamernik et al.¹⁴

typically reach the asymptotic TS level (ATS) produced by a similar impact noise presented on an uninterrupted schedule over 5 days. The TS estimated with brain stem evoked potentials at 8.0 kHz is shown in Figure 12-8a. Despite the initial TS of around 40 dB there is no indication of any TS recovery over the 5 days of the exposure. Except for smaller ATS at 8.0 kHz, the TS function behaved similarly to the TS function measured by Saunders et al.⁶ (Figure 12-3a). Using the same broadband impact at the same 113 dB peak SPL and 1/s repetition rate, but on a 6 h/d schedule over 20 days, in the chinchilla Hamernik et al.¹⁴ showed a clear TS recovery at 8.0 kHz of about 30 dB. As seen from their data replotted in Figure 12-8b, most of this TS recovery took place during the first 5 days of the exposure. A summary of this data at various levels for the 0.5, 2.0, and 8.0 kHz test frequencies is shown in Figure 12-9. This figure illustrates the group mean recovery or toughening fraction, TS_r/TS_1 , where $TS_r = TS_1 - TS_{16-20}$. The toughening is seen to be greatest for the 8.0 kHz test frequency at the 113 dB peak SPL exposure

condition and decreases as the impact intensity increases. In general, the Hamernik et al.¹⁴ data showed that mean PTS as well as outer hair cell loss decreased, as shown in Figure 12-10, as TS_r/TS_1 increased, thus confirming that the toughening produced by the interrupted exposure paradigm resulted in a reduction in trauma. Also, when the results of these interrupted exposures were compared

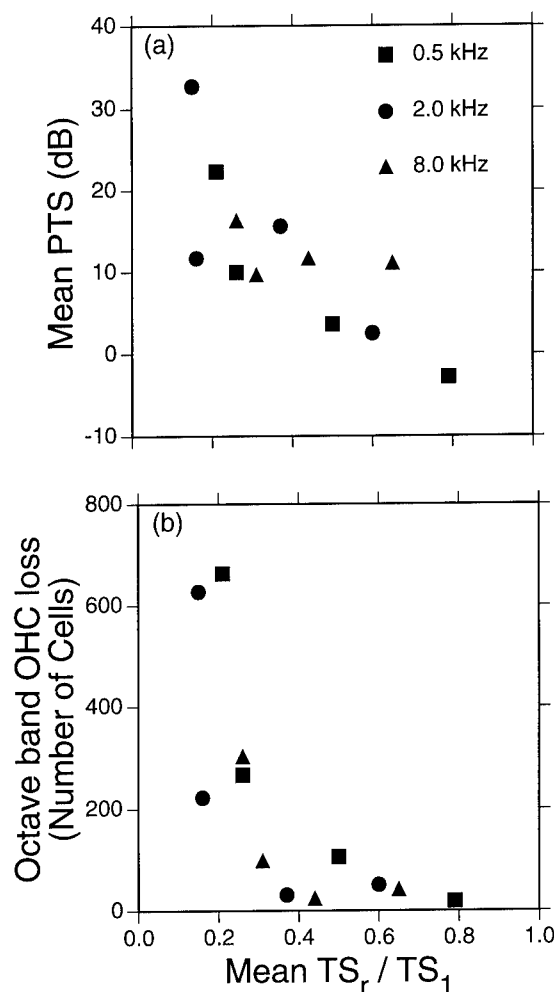


Figure 12-10 Relation between (a) mean permanent threshold shift (PTS) and (b) outer hair cell (OHC) loss in an octave band length of the cochlea and the mean threshold shift recovery function, TS_r/TS_1 , for the 107, 113, 119, and 125 dB peak SPL interrupted impact noise exposures at the indicated test frequencies. From Hamernik et al.¹⁴

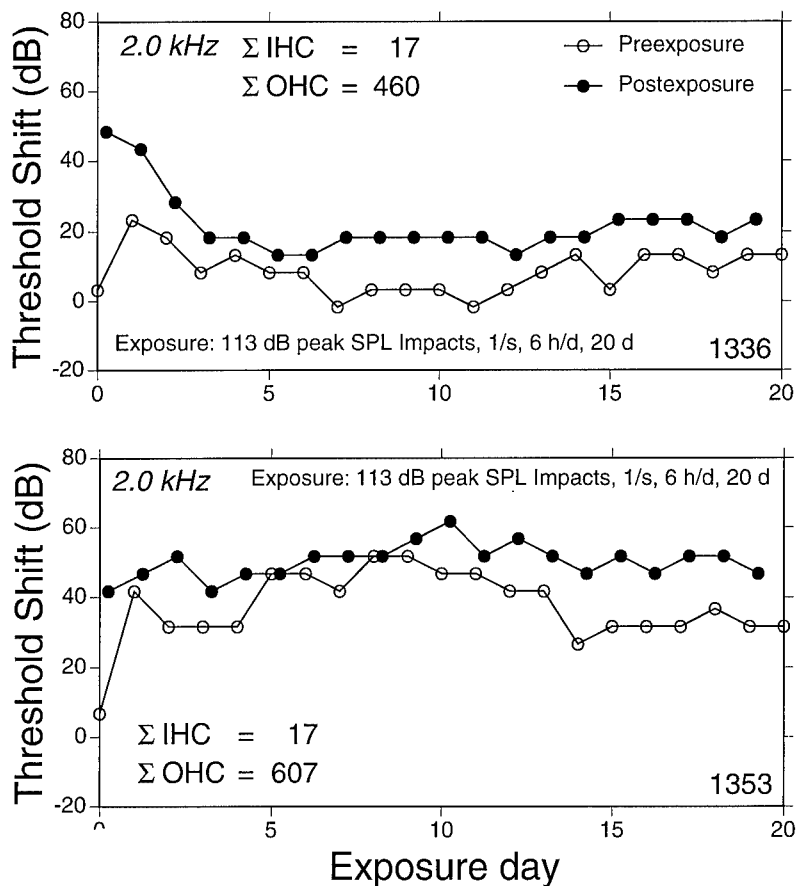


Figure 12-11 The threshold shift measured in Chinchillas 1336 (top) and 1353 (bottom) at 2.0 kHz following daily exposures to 113 dB peak SPL impacts, 6h/d over 20 days. Also recorded are the total number of missing outer and inner hair cells (OHC, IHC) in each of these animals.

with the PTS and sensory cell losses produced by uninterrupted exposures having equal energy, the former were shown to be less traumatic. Although the mean group TS behavior exhibited a reasonable order, there were considerable differences in the TS dynamics across animals. The two examples shown in Figure 12-11 illustrate the two extremes: an animal (1336) that showed a systematic recovery of TS over the first 5 days of exposure and an animal (1353) exposed to the same 113 dB peak SPL impact that exhibited a relatively stable TS. The initial TS was similar in both animals. However, the total sensory cell loss in the octave band length of the cochlea cen-

tered at 2.0 kHz is about 25% greater in animal 1353.

More recently we have begun to use narrowband impacts in an interrupted exposure protocol to study the frequency and intensity manifestations of the toughening effect for noise stimuli that probe the extent of the cochlea. The impact stimuli for the 1.0 and 4.0 kHz conditions along with their spectra are shown in Figure 12-12. The impacts were presented at 115 dB peak SPL, 1/s for 6 h/d over 20 consecutive days. The TS dynamics for both these conditions are shown in Figure 12-13 and 12-14. For the 1.0 kHz impact, there is a clear TS recovery at all but the 16.0 kHz test

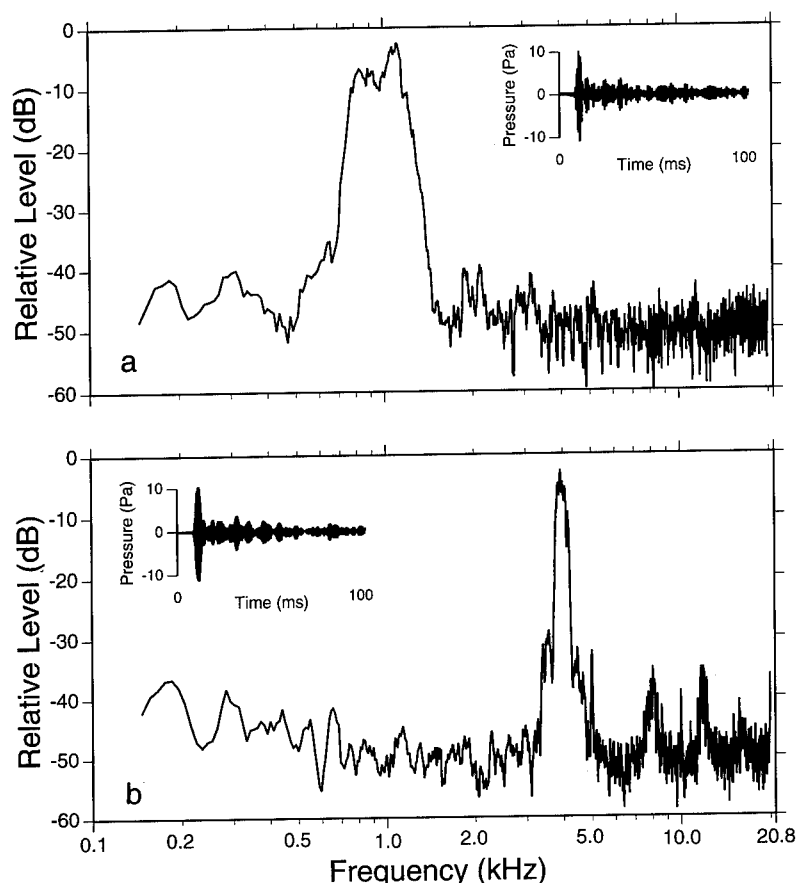


Figure 12-12 The pressure-time waveforms and spectra of the (a) 1.0 kHz and (b) 4.0 kHz narrowband impacts used to study the effect of stimulus frequency on threshold shift recovery functions during interrupted exposure schedules.

frequency where no effect of the noise on the threshold could be recorded with the evoked potential. The maximum toughening effect of about 30 dB occurred at 4.0 kHz, two octaves above the stimulating impact center frequency. For the 4.0 kHz impact, the initial TSs were about the same as for the 1.0 kHz impact, but the TS recovery was relatively small amounting to roughly 15 dB at the 16.0 kHz test frequency, again about two octaves above the stimulating impact. Although there is a slight (<10 dB) recovery at the 4.0 kHz test frequency, there is no TS recovery an octave above or below. This result is similar to the lack of TS recovery seen in the Saunders et al.⁶

results using a 4.0 kHz OBN, but is at odds with the results of Subramaniam et al.¹¹ The latter results shown in Figure 12-6 indicate more than 20 dB TS recovery at 4.0 and 8.0 kHz from a 4.0 kHz cf OBN. The most parsimonious explanation for these discrepancies may lie in the different responses of individual animals. For example, Figure 12-15 shows two animals from the 4.0 kHz narrowband impact study; animal 1929 shows a clear TS recovery at 4.0 through 16.0 kHz and animal 1940 shows no TS recovery at these same frequencies. It is possible that by random sampling a group of animals may reflect one extreme or the other. Animals such as 1929 may

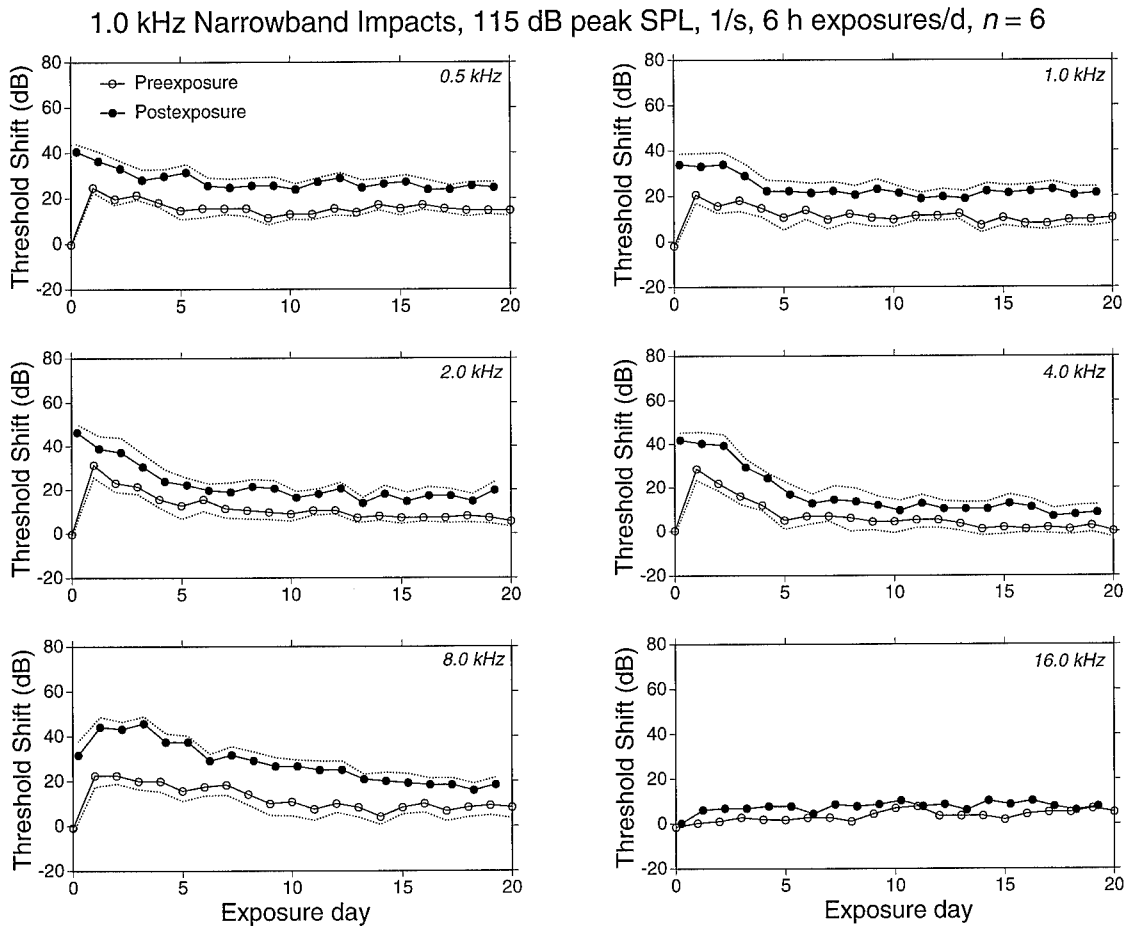


Figure 12-13 Group mean threshold shifts prior to and following each daily exposure to the 1.0 kHz narrowband impacts at the indicated test frequencies (dotted lines indicate \pm one standard error of the mean).

represent a more resistant subpopulation and animal 1940 may be from a subpopulation more susceptible to noise trauma.

3DPEs ($2f_1 - f_2$) were collected using the Entymotic Research CUB^eDIS[™] (version 2.40) system. 3DPEs were collected from each of the six subjects exposed to the interrupted 1.0 kHz narrowband impact. The parameters of the 3DPE collection from which isoemission contours were obtained were: $873 \leq f \leq 9062$ kHz, where $f = \sqrt{f_1 f_2}$; $f_2/f_1 = 1.22$; $L(f_1)$ dB SPL = $L(f_2)$ dB SPL, $20 \leq L \leq 70$ dB SPL in 10 dB steps; 32 (3DPE) points/octave, with an averaging time of 2 s/point. From these high-

resolution 3DPE amplitude-frequency functions (DPEgrams), a set of (at most) 6-point input-output (IO) functions (32 IO functions/octave) were obtained. A routine was developed to extract isoamplitude 3DPEs from this data set. Only 3DPE data points that exceeded the noise floor by 5 dB were accepted as valid in the construction of the isoamplitude functions (IAF). Figure 12-16a illustrates the mean 10 dB IAF for the group exposed to the interrupted 1.0 kHz narrowband impact. The dotted line shows the preexposure 10 dB IAF and the heavy solid line the 10 dB IAF measured after the first 6 hours of

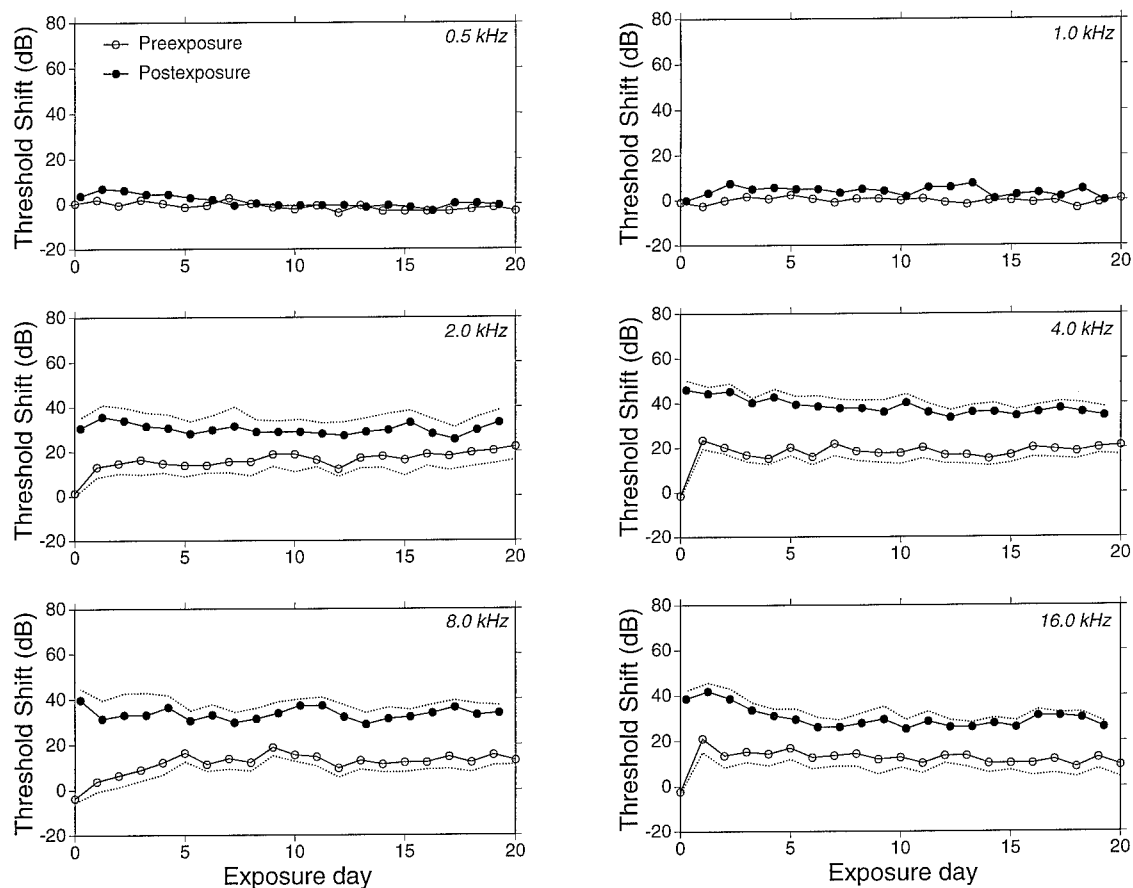
4.0 kHz Narrowband Impacts, 115 dB peak SPL, 1/s, 6 h exposures/d, $n = 6$ 

Figure 12-14 Group mean threshold shifts prior to and following each daily exposure to the 4.0 kHz narrowband impacts at the indicated test frequencies (dotted lines indicate \pm one standard error of the mean).

exposure. The IAF is shifted over 20 dB at most frequencies and for frequencies around 1.0 and 4.0 kHz could not be recorded because of even greater shifts. The lighter solid line shows the IAF measured immediately after day 20 of exposure. Although still elevated across the frequency range of measurement, there is considerable recovery, especially in the 1.0 and 4.0 kHz regions. In general, there is a fair congruence at the corresponding test frequencies between the amount of TS shift following day 1 and day 20 and the shift in the primary levels needed to obtain the 10 dB IAF. The recovery of 3DPEs can also be seen in DPEgrams shown in Figure 12-16b. The noise

floors for these data are not shown because the data presented in the figure are above the noise floor.

In conclusion, the results of interrupted noise exposures show that consistent recovery of threshold can be measured despite the continuing exposure. The phenomenon is repeatable; is found with both impact and continuous noise exposure paradigms; exhibits frequency specificity; is demonstrable within a limited dynamic range of exposure intensities; and can be measured by various experimental protocols from VIII nerve single-unit recordings, to behavioral thresholds and otoacoustic emissions. The implications for the

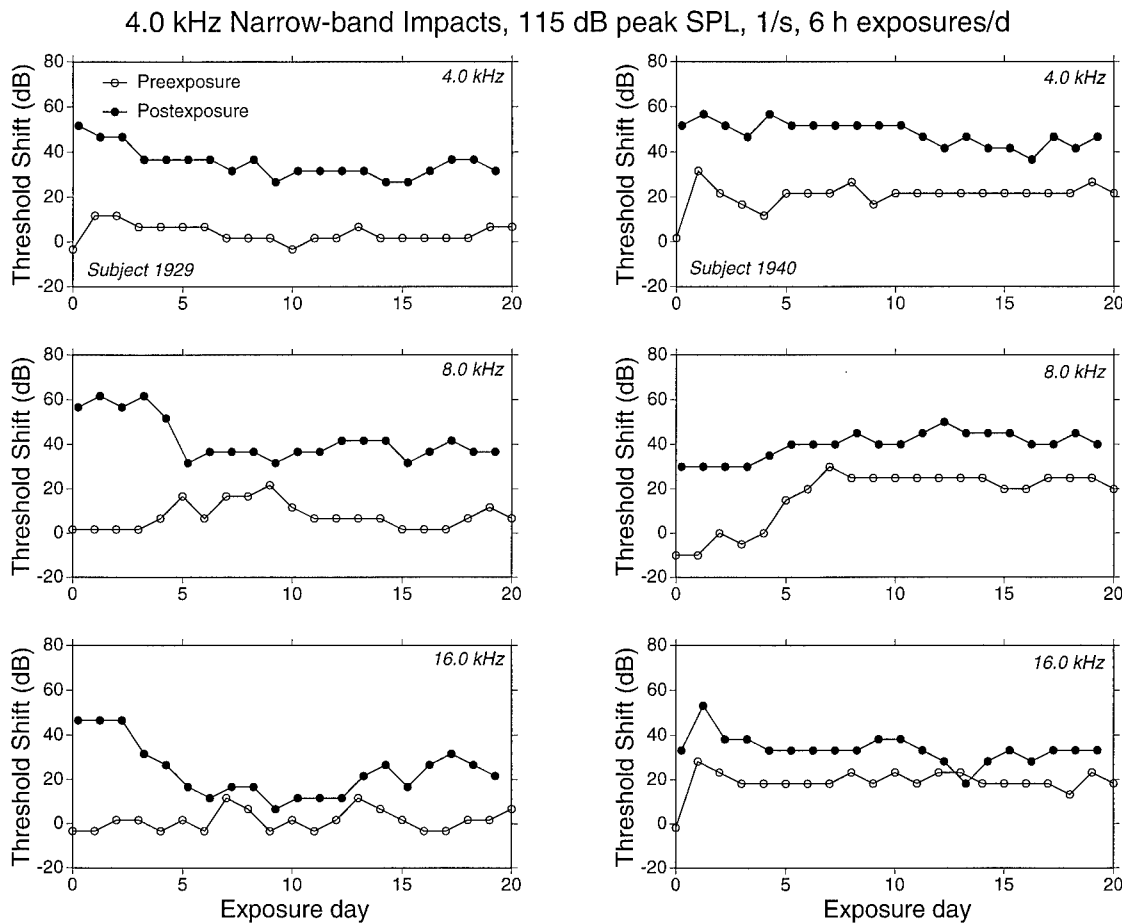


Figure 12-15 Threshold shifts prior to and following each daily exposure to the 4.0 kHz narrowband impacts for two individual animals (1929 and 1940) at the indicated test frequencies.

reduction of permanent hearing loss are unclear given the often conflicting relationship between PTS and sensory cell loss. The considerable differing TS dynamics across animals and the relation of these TSs to PTS and cell loss data is also problematic. The TS recovery phenomenon further illustrates the inadequacy of the equal-energy approach to noise evaluation for hearing conservation purposes.

Acknowledgments

This work was supported by Grant 1-R01-OH02317 from the National Institute for Occu-

pational Safety and Health. The able technical assistance of C. Case and clerical skill of P. Bridges are greatly appreciated.

Animal Use

In conducting the research described in this study, the investigators adhered to the *Guide for the Care and Use of Laboratory Animals* prepared by the Committee on Care and Use of Laboratory Animals of the Institute of Laboratory Animal Resources, National Research Council [DHHS Publication No. (NIH) 86-23, revised 1985].

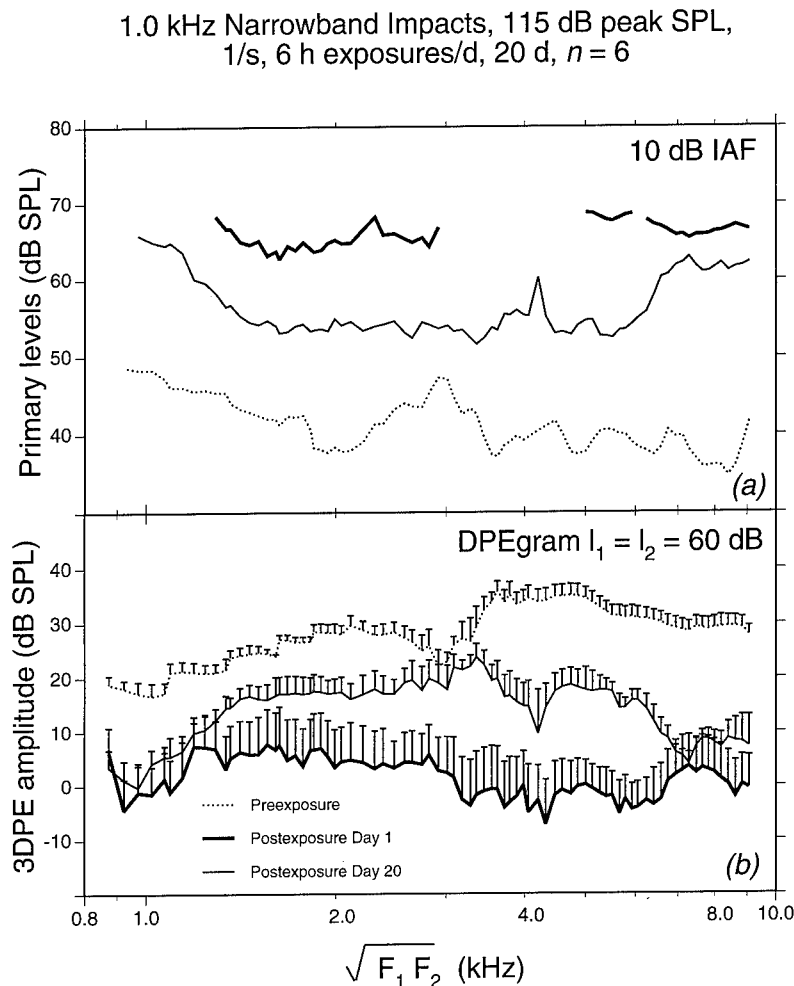


Figure 12-16 (a) The mean level of the primary tones required to generate a 10 dB SPL cubic distortion product otoacoustic emission (3DPE) and (b) the mean level of the cubic distortion product otoacoustic emission for a 60 dB SPL primary tone at various frequencies prior to exposure to an interrupted 1.0 kHz narrow band impact and following the first day of exposure and the last day (day 20) of the exposure. (Error bars represent one standard deviation.)

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Chapter 13

Protection from Continuous, Impact, or Impulse Noise Provided by Prior Exposure to Low-Level Noise

Donald Henderson, Malini Subramaniam,
Lynn W. Henselman, Paola Portalatini,
Vlasta P. Spongr, and Vincenzo Sallustio

Canlon et al.¹ showed that the resistance of the auditory system to noise-induced hearing loss (NIHL), could be increased with a prior prophylactic or “conditioning” exposure. In their experiments, guinea pigs that were exposed to a low level (81 dB), 1 kHz tone for 24 days followed by an exposure to the same tone at 105 dB sound pressure level (SPL) for 72 hours, developed 20–25 dB less permanent threshold shifts (PTSs), than a control group only exposed at the higher level. The size of the protective effect is very large (15–30 dB) and raises questions about the biological mechanisms of the phenomenon as well as the appropriate acoustic parameters that generate an increase in resistance to noise. As a first step in understanding the “toughening” phenomenon, our laboratory has begun a program of experiments designed to understand the acoustic parameters governing the phenomenon of acquired resistance or toughening, that is, the spectrum of the conditioning noise, the duration of conditioning exposure, the persistence of the protective effect, and finally, the effect of conditioning exposures on the mechanical trauma caused by impact/ impulse noise.

Method

Monastral chinchillas served as subjects in all the experiments to be described. Their hear-

ing sensitivity was assessed using evoked potential (EVP) recordings made from a chronically implanted electrode in the contralateral inferior colliculus.² A standard experimental protocol was used for each of the experiments (Figure 13-1). First, the subject's preexposure thresholds were measured five times, at octave frequencies from 0.5 to 8 kHz, and the average of the five measures served as the baseline for the given frequency. This preexposure baseline was used as the reference for calculating temporary threshold shifts (TTSs) and PTSs. After the preexposure tests, subjects were exposed for 10 days to a lower level (conditioning), octave band of noise (OBN). EVP thresholds were measured just before and after each day's exposure. After the last day of conditioning exposure, subjects were allowed to recover for a 5 day period and then reexposed to a higher level noise. PTSs were measured 30 days after the high-level exposure and the results were compared with a control group given only the corresponding higher level exposure.

Results

Toughening at Low Frequencies

In the Canlon et al.¹ experiment a 1 kHz tone at 81 dB for 24 days was used as the condition-

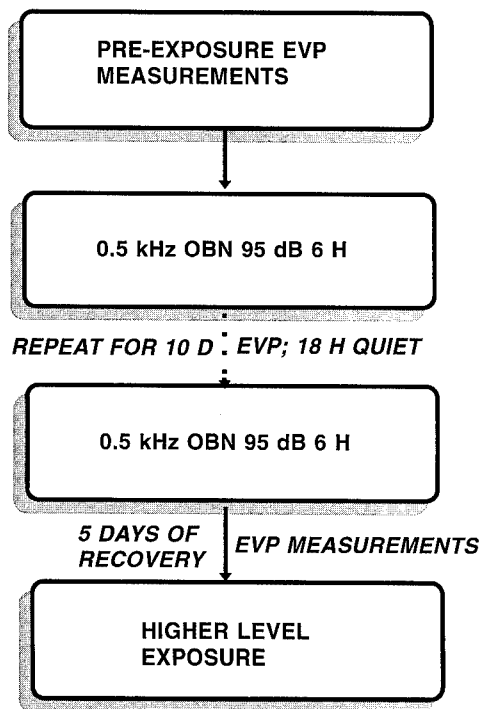
EXPOSURE SCHEDULE

Figure 13-1 Schematic diagram of the experimental schedule in the experiments on toughening.

ing exposure and thresholds were measured using surface electrodes in guinea pigs. This exposure has little in common with the acoustical characteristics of industrial noise. In our first experiment³ the goal was to see whether a low-frequency exposure could produce a similar protective effect. Thus, the conditioning exposures consisted of an OBN centered at 0.5 kHz presented at 95 dB SPL for the 10 days. After a 5 day recovery period, the subjects were reexposed to the same noise at 106 dB for 48 hours. Figure 13-2 shows the average PTS of six experimental subjects compared against the PTS in six control subjects. The PTS in the experimental group is 10–15 dB lower than in the control group and these differences were statistically significant ($p < 0.05$). These results are also consistent with Canlon's (1 kHz exposure) and show that the toughening phenomenon is present in at least two separate species.

Number of Conditioning Exposures

The initial experiments by Canlon et al.¹ and Campo et al.³ in our lab, used a relatively long duration conditioning exposure (24 days continuously; 10 days for 6 h/d, respectively). A reasonable question to ask is whether the protective effect produced by the series of 10 day exposures is either increased with more exposures or diminished with fewer exposures. To answer these questions, three experimental groups of chinchillas were exposed to either 6 h/d for 20 days, 6 h/d for 10 days, or two 6 hour exposures separated by 8 days of quiet.⁴ After each of the conditioning exposures, the subjects were maintained in a quiet environment for 5 days. Figure 13-3 shows that all three experimental groups developed significantly less PTS than the control group. Of the experimental groups, it is interesting to note, that the 20 day group had the largest amount of PTS. Although the differences are not significant, they suggest that prolonged exposures at the level used in this experiment (95 dB), may actually decrease the prophylactic effect. Conversely, the 2 days of exposure produced almost as much protection as 10 days. This finding may be important for the eventual application of the toughening phenomenon because it shows that the increased resistance can be created with a minimal investment in time. Collectively, these results focus on a practical application of the toughening phenomenon, namely, the minimal duration and level of the conditioning exposure that could produce a significant increase in resistance to noise.

Persistence of Toughening

In the above experiments, the acquired resistance was seen 5 days after the series of conditioning exposures. If the toughening phenomenon is to be of possible practical significance, then it is important to understand whether the toughening effects are persistent for longer periods of time. In a recent experiment in our lab, subjects were given the 10 days of prophylactic exposures and then were kept in a quiet animal colony for 30 days. Figure 13-4

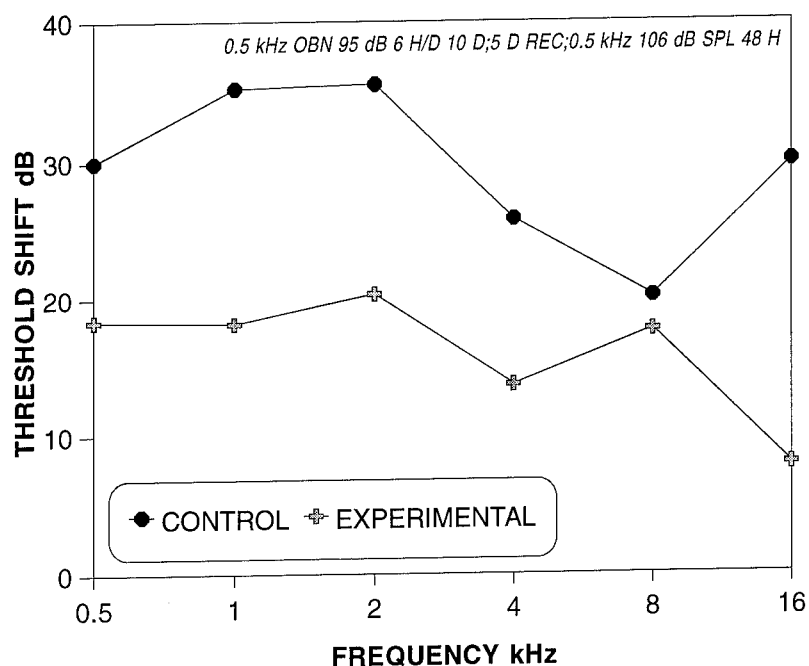


Figure 13-2 Effect of conditioning exposures (OBN centered at 0.5 kHz at 95 dB for 6 h/d for 10 days) on PTS from a higher level exposure (106 dB SPL). The experimental group developed lower PTS (10 to 15 dB) than did the control group exposed only at the higher level.

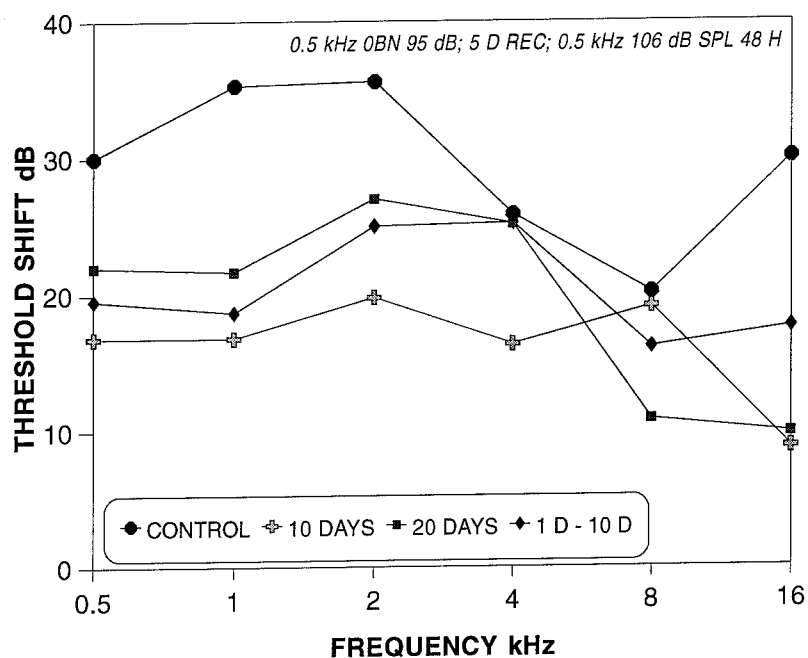


Figure 13-3 Effect of number of conditioning exposures on PTS. Irrespective of the number of conditioning exposures, all the subjects that were previously exposed at a lower level incurred lower PTS than the control subjects following the 106 dB exposure for 48 hours.

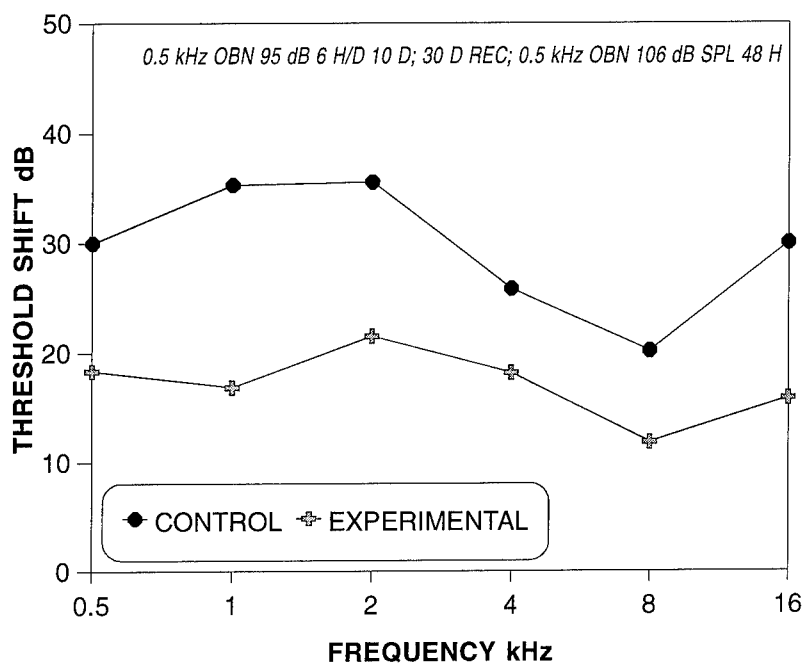


Figure 13-4 Persistence of the conditioning effect. Animals that received prior conditioning (OBN centered at 0.5 kHz at 95 dB for 6 h/d for 10 days) incurred lower PTS than did the control subjects even after a lapse of 30 days between the conditioning and the higher level exposures at 106 dB SPL for 48 hours.

shows that the experimental subjects developed 10–15 dB less PTS than the control subjects, despite a 30 day of lapse time between the conditioning and the higher level exposures.

The fact that the auditory system maintains its increased resistance for 30 days raises interesting questions about what the biological basis of toughening might be. For example, it is highly unlikely that any change to the efferent system or the acoustic reflex would persist for such a long time. Perhaps even more importantly, the persistence of the acquired resistance makes the possibility of using the phenomenon for practical applications more realistic.

Toughening at High Frequencies

Canlon et al.¹ and Campo et al.³ have shown that the toughening phenomenon works for low-frequency conditioning followed by low-

frequency “traumatic” exposures. This raises the question of the generality of the phenomenon across the frequency spectrum. Do low-frequency conditioning exposures protect against future high-frequency traumatic exposures? Do high-frequency conditioning exposures produce progressively less TS with repeated exposures? Finally, do high-frequency conditioning exposures protect against future high-frequency traumatic exposures?

Figures 13-5, 13-6, and 13-7 provide some answers to the above questions. Figure 13-5 shows the average daily threshold shifts following 6 hour exposures to an OBN centered at 4 kHz at 85 dB SPL. There is a marked decrease in TSs as the number of daily exposures increases.⁵ In fact, when the TSs at 1 octave above the exposure frequency are compared, the TS is seen to reduce faster with the high-frequency exposure than with a similar low-frequency (OBN centered at 0.5 kHz) exposure. These results are consistent with experiments by Clark and Bohne.⁶

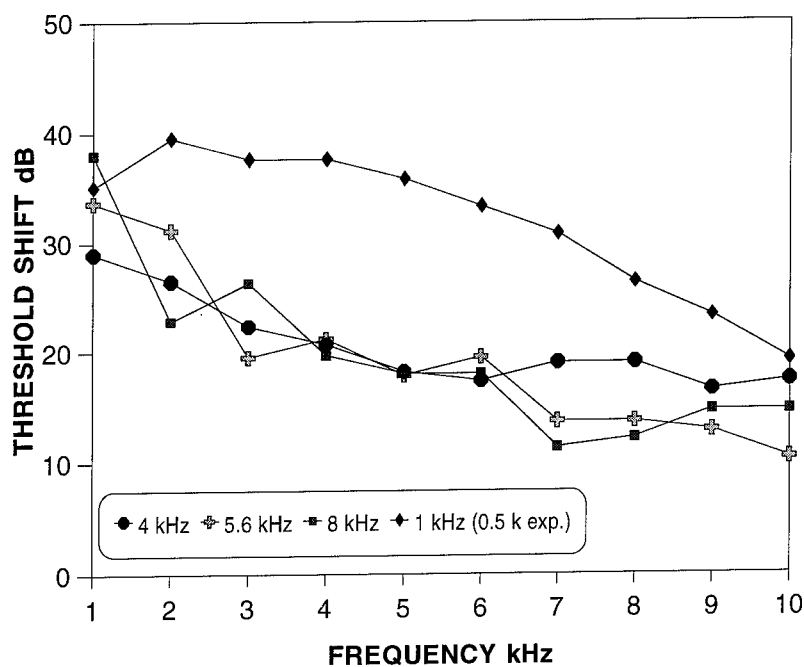


Figure 13-5 TS from exposures to an OBN centered at 4 kHz at 85 dB SPL for 6 h/d for 10 days. Note the large reductions in TS at all test frequencies. Comparison with the reductions in TS at 1 kHz following a low-frequency exposure (OBN centered at 0.5 kHz, at 95 dB SPL) shows similar initial and final TS, but a more rapid reduction in TS following the high-frequency exposure.

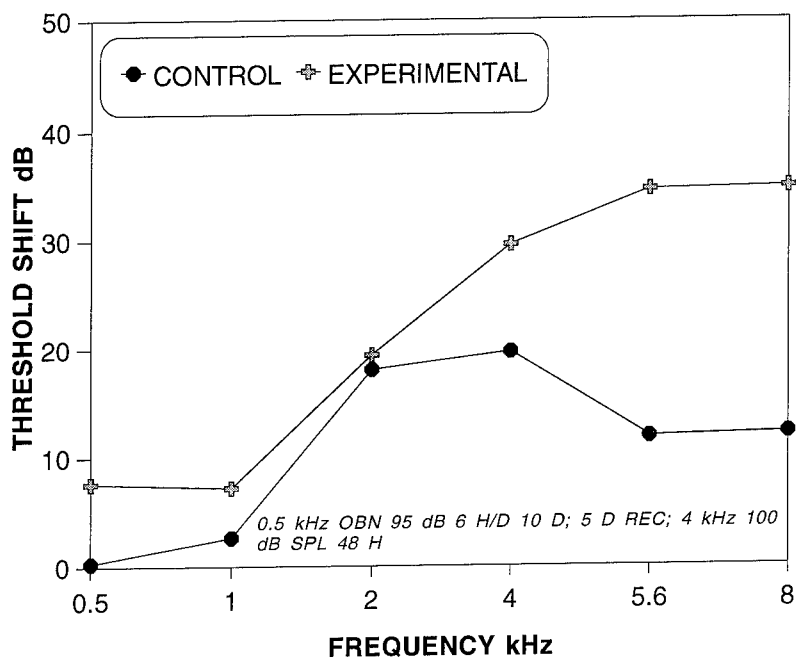


Figure 13-6 Effect of low-frequency conditioning (OBN centered at 0.5 kHz at 95 dB for 6 h/d for 10 days) exposures on PTS from a higher level exposure, at a high frequency (OBN centered at 4 kHz at 100 dB SPL for 48 hours). The experimental group developed greater PTS (about 20 dB) than did the control group exposed only at the higher level (100 dB SPL).

In spite of the reduced TS with repeated exposures, the effects of high-frequency conditioning on PTS are not as encouraging or clear. When subjects are given the 10 days of low-frequency conditioning (OBN centered at 0.5 kHz, 95 dB SPL) and then followed 5 days later with an exposure to an OBN centered at 4 kHz at 100 dB SPL, experimental subjects develop significantly more PTS than the control group⁷ (Figure 13-6). Furthermore, if the conditioning exposure is changed to 4 kHz at 85 dB and the traumatic exposure (OBN centered at 4 kHz, 100 dB SPL) is followed 5 days later, the subjects develop more PTS than the control group (Figure 13-7). However, if the traumatic exposure is presented 18 hours after the last of the conditioning exposure, the subjects develop less PTS than a control group⁸ (Figure 13-7).

In summary, it appears that toughening or acquired resistance may work differently with basal and apical regions of the cochlea. Both

high- and low-frequency conditioning exposures show a robust decrease in TS with repeated exposures. However, the long-term protective effects are either less persistent or even nonexistent for the high-frequency region of the cochlea and the conditioning effect may have the opposite effect and actually render the ear more susceptible to future high-frequency noise-induced trauma.

Impulse and Impact Noise

Impulse and impact noise present a special hazard to the cochlea. The high peak amplitudes and rapid rise times associated with these types of noise, damage the cochlea by causing mechanical failure such as ripping of the organ of Corti off the basilar membrane, separating the OHC from the Hensen's cells, and rupturing tight cell junctions at the reticular lamina^{9,10} (Figure 13-8). Recently, our laboratory has begun to investigate the possibility

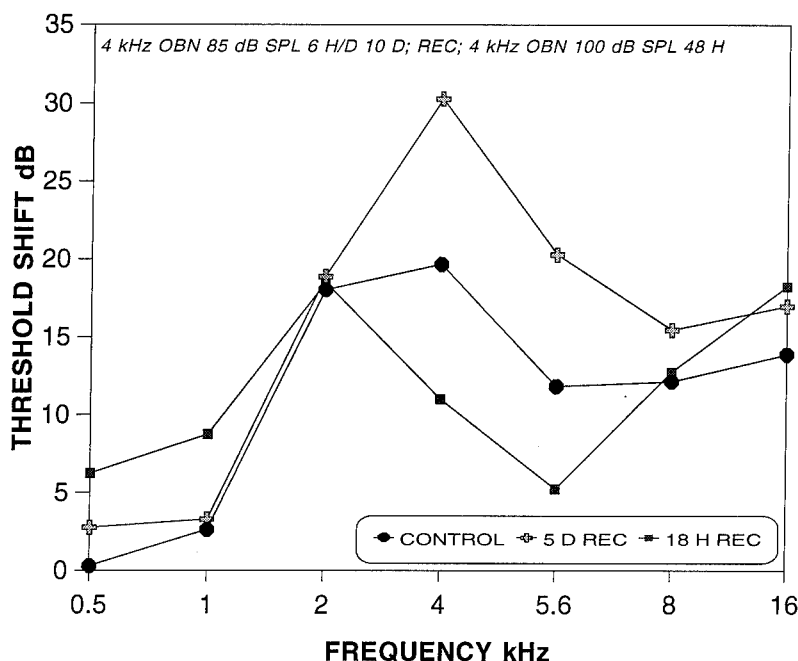


Figure 13-7 Effect of high-frequency conditioning (OBN centered at 4 kHz at 85 dB for 6 h/d for 10 days) exposures on PTS from exposures to an OBN centered at 4 kHz at 100 dB SPL for 48 hours. When the experimental group was rested for 5 days before exposure to the OBN at 4 kHz at 100 dB SPL for 48 hours, they developed 10 dB more PTS than the subjects in the control group. However, if the experimental subjects were given only 18 hours between the conditioning and the higher level exposures, they developed PTSs that were significantly lower than in control subjects.



Figure 13-8 Scanning electron microscopical view of the organ of Corti after 24 hours of recovery following an exposure to an impact noise at 137 dB. Note the ripping and swirling of the organ of Corti and Hensen cells off the basilar membrane (photomicrograph from 0.8 kHz region). Remaining outer hair cells and Deiters cells appear grossly degenerated.

of applying the toughening phenomenon to impulse and impact noise.

Two sets of experiments were conducted. The first experimental design required that the subject be exposed to an OBN centered at 0.5 kHz at 95 dB SPL for 6 h/d for 10 days. After the last of the series of exposures, subjects recovered in a quiet animal colony for 5 days, then retested, and then exposed to impulse noise that mimicked a M-16 US Army rifle. The peak level of the noise was 150 dB and the subjects were exposed to 50 pairs of impulses separated by 1 second (Figure 13-9). Figure 13-10 compares the PTS 30 days after the exposure with a control group only exposed to the impulse noise. The subjects with the prior conditioning exposures developed substantially less PTS. Furthermore, compari-

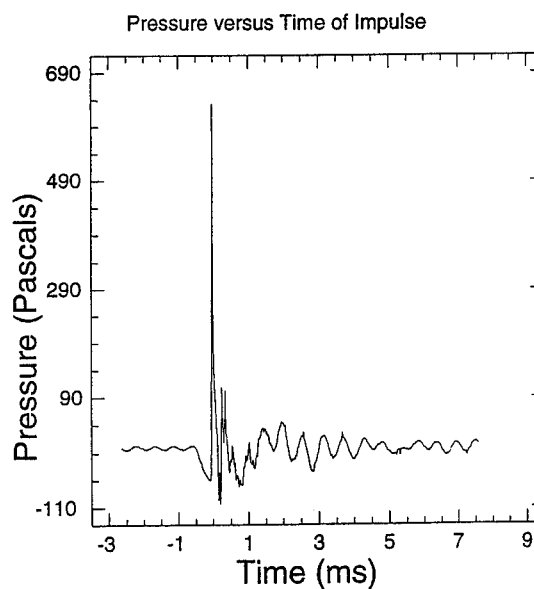


Figure 13-9 Acoustic wave form of the impulse, mimicking an impulse from a M-16 rifle.

sons of average cochleograms revealed that the cochleas from the control group were much more extensively damaged than those from the experimental group.¹¹

The reduced PTS following exposure to impulse noise is particularly interesting considering that soldiers are often exposed to high level noise without the benefits of a hearing protection device. Given that the chinchilla can be made more resistant to noise with only two, six hour exposures and that the protective effect lasts for at least one month, then, it is reasonable to consider the possibility of protective exposures for certain military personnel.

In the second experiment, the experimental schedule and conditioning exposures were the same except that the traumatic exposure was to an impact noise with a peak level of 131 dB and a duration of 200 milliseconds presented at the rate of 1/s for 1.8 hours. The average PTS was significantly less (10–15 dB at 0.5 and 1 kHz) in the experimental group (Figure 13-11) and the cochlear damage was slightly more severe in the control group. The prophylactic effect for impact noise was significant as reflected in lower PTS and hair cell

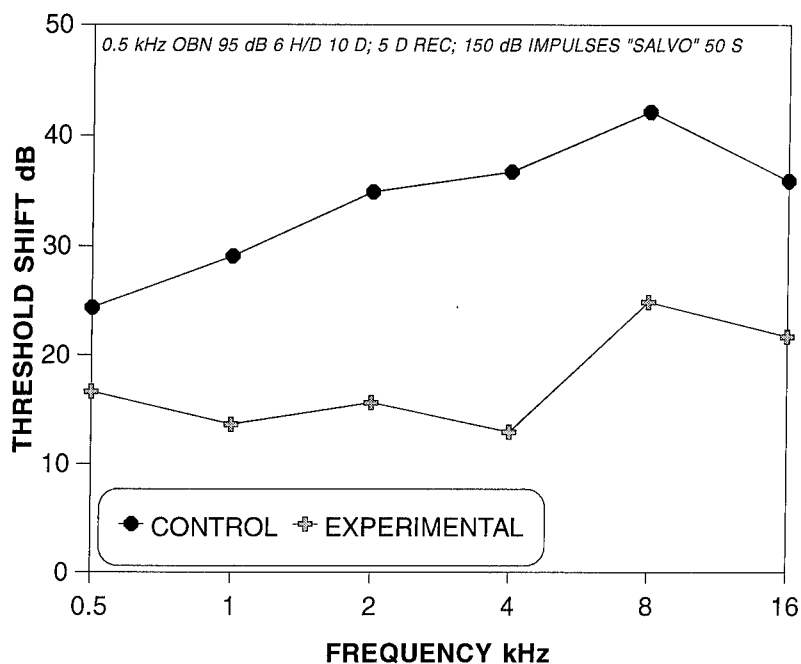


Figure 13-10 Effect of conditioning (OBN centered at 0.5 kHz at 95 dB for 6 h/d for 10 days) on PTS from exposure to impulse noise. Animals subjected to low-frequency conditioning exposures developed significantly lower PTS than did the control subjects at all test frequencies after exposure to impulse noise at 155 dB.

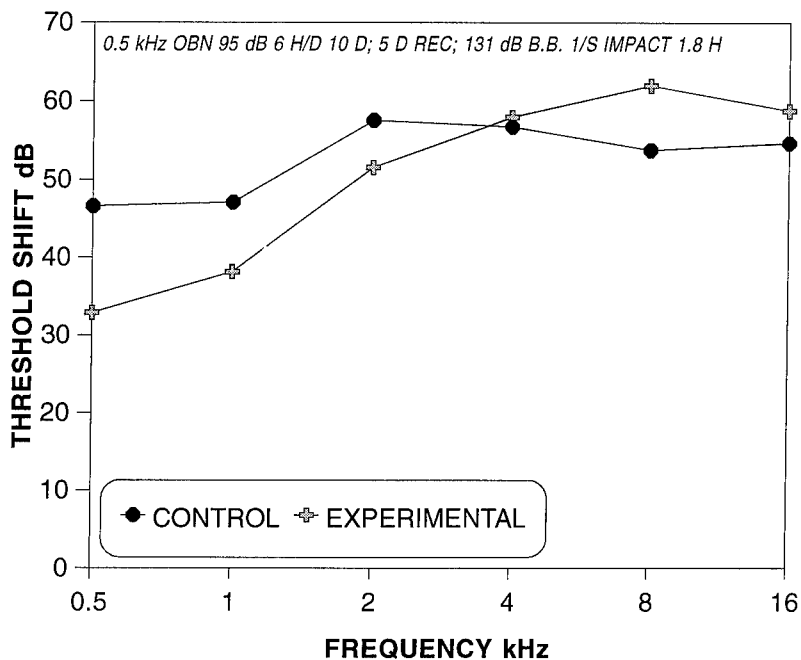


Figure 13-11 Effect of conditioning (OBN centered at 0.5 kHz at 95 dB for 6 h/d for 10 days) on PTS from exposure to impact noise at 131 dB. Animals subjected to low-frequency conditioning exposures developed significantly lower PTS than did the control subjects at low frequencies.

loss, but the effect was less pronounced than for the impulse noise.

Summary

The above series of experiments help to "map" out the range of acoustic parameters associated with the toughening phenomenon. The initial demonstration¹ of toughening involved a low-frequency sinusoid. Our work³ has replicated this study with low-frequency noise. The toughening effect seems to be established with relatively short duration prophylactic exposures: as few as two, 6 h/d exposures.⁴ The prophylactic effect is persistent and there is clear evidence of increased resistance up to 30 days after the prophylactic exposure. By contrast, the toughening effect does not seem to be significant at high frequencies, with the possible exception of an increased resistance immediately after the exposure that dissipates quickly.⁸ Noteworthy are the results of two experiments that provide strong evidence for protection from the mechanical damage associated with either impulse or impact noise.¹¹

The toughening phenomenon raises interesting basic science questions and suggests the possibility of actually developing a prophylactic procedure for people that are exposed to high levels of noise, but for special reasons cannot use personal protection devices.

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Chapter 14

Efferent and Priming Modulation of Noise-Induced Hearing Loss

Ramesh Rajan

Noise-induced hearing losses (NIHLs) at the mammalian cochlea can be modulated in a variety of ways. Most consistently it has been shown that NIHL can be modulated by activating the efferent pathways to the mammalian cochlea or those to the middle ear muscles (MEMs), and by "conditioning" with sound. Such modulatory effects are obviously of great interest both scientifically, in the study of cochlear mechanisms, and clinically, in terms of predicting and ameliorating the damage produced by loud sounds. The role of the MEMs in reducing the damage caused by loud sounds has been reviewed elsewhere (e.g., Borg et al.¹) and the features of the modulatory effects of conditioning on NIHL are presented in detail in other chapters in this book. Hence this chapter will be concerned primarily with the modulatory effects of the efferent pathways to the mammalian cochlea. Initially I briefly review the major features of the modulatory effects of the cochlear efferent pathways on NIHL described in previous studies in one species before presenting details of more recent data from another species. Finally, I discuss the interaction between the modulatory effects of these efferent pathways and the modulatory effects of short-term sound conditioning.

Olivocochlear Bundle-Mediated Protection from NIHL in Guinea Pigs

It has now been clearly established in a number of studies in the guinea pig²⁻¹⁶ that NIHL at the cochlea can be significantly reduced by

activation of the olivocochlear efferent pathways. Detailed examination in at least some of these studies suggest that the protective effects can be attributed specifically to the crossed olivocochlear pathways²⁻⁶ originating from nuclei located mediolaterally in the superior olivary complex, and the cochlear effects appear almost certainly to be exercised at the outer hair cells via nicotinic cholinergic synapses.⁷ Thus, this protective role appears to be carried out only by this one subcomponent of the olivocochlear pathways, namely, the nicotinic cholinergic synapses of the crossed olivocochlear pathways.

Despite early reports to the contrary,^{17,18} more recent studies have found that similar olivocochlear bundle-mediated protection from NIHL can also be obtained in the cat, as will be demonstrated here. Another recent study¹⁵ has also found similar protective effects from NIHL in the rat. Collectively these data confirm the generality of the protective function of at least one component of the olivocochlear pathways.

The studies in guinea pigs, particularly my own studies, have been reviewed in detail previously.^{19,20} Hence only a brief summary of the major features of these studies will be presented here. This review will serve merely to provide a framework to discuss more recent data obtained from studies in the cat on the modulatory effects of the olivocochlear pathways.

The basic modulatory effect of the olivocochlear bundle (OCB) on NIHL revealed in the guinea pig studies is a protective one to

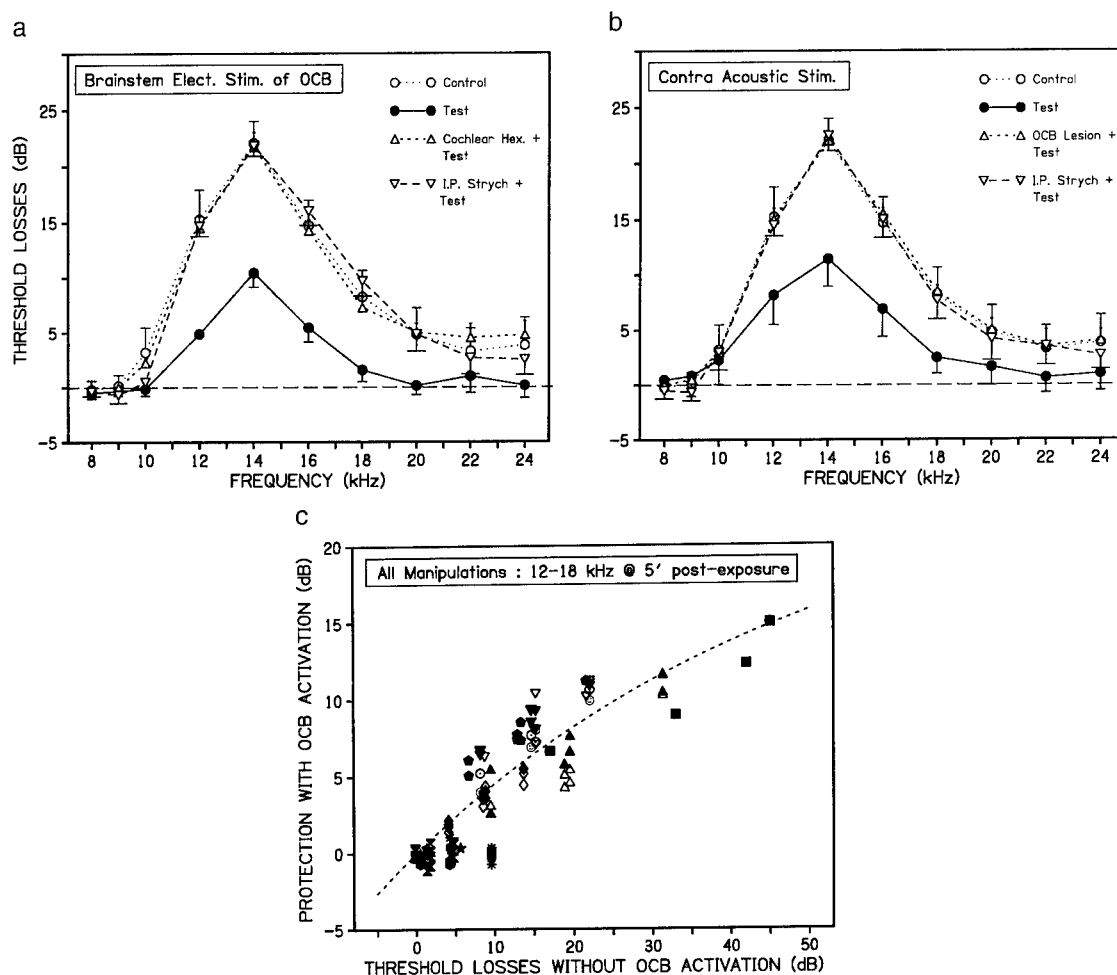


Figure 14-1 Basic features of the protective effects of the olivocochlear pathways in the guinea pig. Panels **a** and **b** show the protective effects of electrical stimulation at the floor of the fourth ventricle (**a**) or of contralateral acoustic stimulation (**b**), and data from tests to show that these were due to the OCB. The data are mean threshold losses (error bars = SEM) 5 minutes postexposure in each group of animals. The standard monaural loud sound exposure was always at 10 kHz, 103 dB sound pressure level (SPL) for 1 minute. (**a**) Protection with electrical stimulation of the OCB. Control: standard exposure alone. Test: standard exposure + simultaneous OCB stimulation (bipolar 400 μ A pulses at 140/s). Cochlear Hex + Test: intracochlear perfusion of hexamethonium about 15 minutes prior to test conditions. IP Strych + Test: IP injection of strychnine about 40–60 minutes prior to test. (**b**) Protection with contralateral acoustic stimulation. Control: standard exposure alone. Test: standard exposure + simultaneous stimulation of the contralateral ear at 10 kHz, 80 dB SPL for 1 minute. OCB Lesion + Test: Lesioning of COCB in brain stem midline prior to test conditions. IP Strych + Test: IP strychnine (at 4 mg/kg body weight) about 15 minutes prior to test. (**c**) Relationship between the amount of protection with OCB activation and the amount of loss that would otherwise ensue. Data are based on threshold losses recorded at frequencies from 12 to 18 kHz, 5 minutes postexposure. Abscissa: Mean losses in control groups presented only monaural exposures. Ordinate: amount of protection, measured as the difference at each frequency from 12 to 18 kHz in mean losses in control groups and mean losses in test groups in which the same exposure was combined with a test manipulation. Exposures were always at 10 kHz, and in different groups ranged in intensity from 97 to 110 dB SPL, and in duration from 10 to 60 seconds. Test manipulations were electrical stimulation of the OCB at 140 pulses/s simultaneously, with and for the duration of the exposure, either at the floor of the fourth ventricle (bipolar stimulation), at the round window of the cochlea (bipolar or

reduce the cochlear desensitization caused by loud sound exposures. In these studies^{2-11,16} the NIHL was created using a loud sound exposure at 10 kHz. This is a frequency in the midrange of the guinea pig's hearing range, and one that produces losses in cochlear sensitivity in the region of greatest hearing sensitivity of the species, as assessed by audiograms constructed from the compound action potential recorded directly from the round window of the cochlea. Hearing losses were generally assessed as changes in threshold sensitivities measured using the compound action potential audiogram. The experiments were carried out in guinea pigs anesthetized with pentobarbitone sodium and paralyzed to eliminate the action of the MEMs.

In these studies OCB-mediated protection was elicited by a variety of manipulations, and data from two of these manipulations are illustrated in Figure 14-1. The effects in these studies can be summarized as follows.

1. Direct activation of the OCB^{2,3,9,10} can reduce NIHL. In Figure 14-1 (panel a) in the test group, the standard loud sound exposure was combined with electrical stimulation of the OCB in the brain stem midline at the floor of the fourth ventricle. Significantly lower threshold losses were recorded compared to losses in the control group that presented only the exposure. When strychnine (the classical blocker of the OCB) was injected intraperitoneally (IP) at a dose of 4 mg/kg body weight about 45 minutes prior to the test conditions,^{2,9} the protective effect of the electrical stimulus was completely blocked (see Figure 14-1). As detailed elsewhere,^{2,9} the time course of this block of electrically elicited protection from NIHL was similar to the drug's blocking action on other electrically elicited efferent effects at the cochlea.^{2,9,21}

2. The protective effects at the cochlea are exercised by the nicotinic cholinergic syn-

apses of the OCB.⁷ When the antinicotinic cholinergic agent, hexamethonium, was infused into the cochlea prior to applying the test condition using electrical stimulation (Figure 14-1, panel a), the protective action of the brain stem electrical stimulus was completely blocked. Again, as detailed elsewhere,⁷ the time course of this block from NIHL was similar to the drug's blocking action on other electrically elicited efferent effects at the cochlea.

3. OCB-mediated protection can also be elicited by manipulations at the cochlea contralateral to that presented the loud sound producing the NIHL.^{4-6,8,11,13} One such manipulation is acoustic stimulation with a low-level (nondamaging) sound at the same frequency as the ipsilateral loud sound exposure.^{4,8,11} The contralateral sound reduces the NIHL caused by the ipsilateral exposure (Figure 14-1, panel b), with the amount of protection being very similar to that achieved with direct electrical activation of the OCB. The protective effects of contralateral cochlear manipulations are blocked by IP injections of strychnine prior to the test conditions or by lesioning the crossed OCB in the brain stem.^{4,8,11} The ability of these interventions to block protection elicited by contralateral acoustic stimulation is shown in the same panel in Figure 14-1.

4. Protection can also be elicited by electrical stimulation at the auditory midbrain, either ipsilateral or contralateral to the test cochlea (see Figure 14-2, panel a). Such protection was elicited from the inferior colliculus (IC).⁷ This protection is blocked by intracochlear infusion of hexamethonium (not illustrated here, but see Rajan⁷), confirming that the effects elicited from the midbrain site were exerted specifically through a pathway terminating in the cochlea. As shown in Figure 14-2, at any particular rate of electrical stimulation, greater protection was elicited

monopolar stimulation ipsilaterally, or monopolar stimulation contralaterally), contralateral acoustic stimulation (at 10 kHz, 80 dB SPL) simultaneously with and for the duration of ipsilateral exposure, or contralateral cochlear destruction about 2 minutes prior to ipsilateral exposure. Not all exposures were tested with all test manipulations.

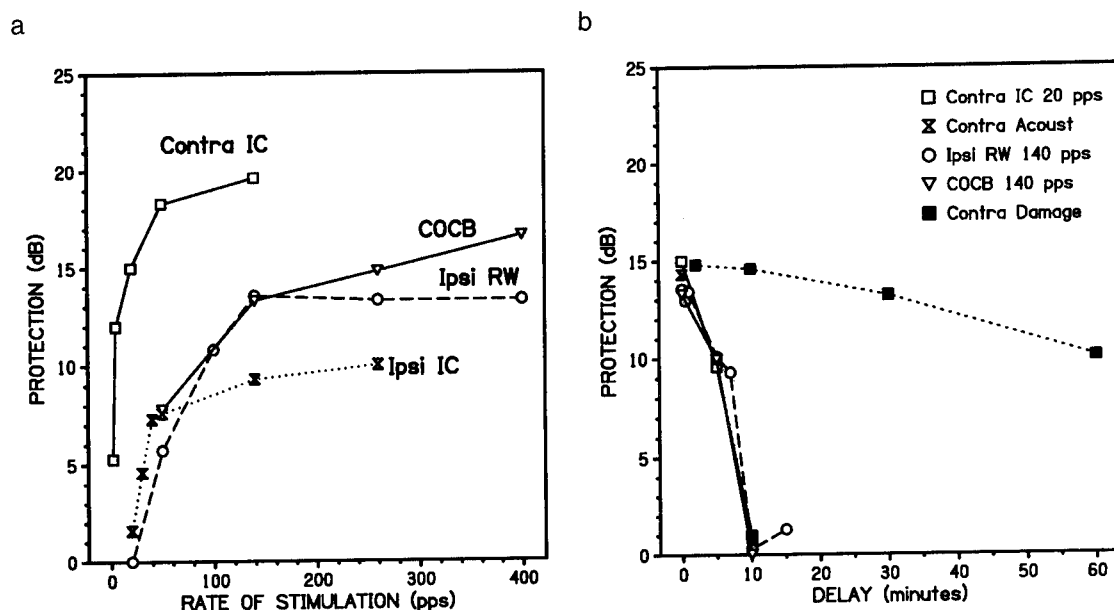


Figure 14-2 Dependency of protection on rate of electrical stimulation (a) or delay between protective manipulation and exposure (b). Protection was measured as the difference in mean peak loss (at 14 kHz, 10 seconds postexposure) between the test group (manipulation + exposure) and the control group (exposure alone). The exposure was always at 10 kHz, 103 dB SPL for 1 minute. (Panel a) Dependency of protection on rate of electrical stimulation. The electrical stimulus was simultaneous with and for the exposure duration. Only the rate was varied at each stimulation site: the inferior colliculus contralateral to the test cochlea (Contra IC); the brain stem midline at the floor of the fourth ventricle (COCB); the round window of the test cochlea (Ipsi RW); or the inferior colliculus ipsilateral to the test cochlea (Ipsi IC). (Panel b) Dependency of protection on delay between test manipulation and standard exposure. The manipulations were: electrical stimulation at 20 pulses/s of the IC contralateral to the test cochlea (Contra IC 20 pps); acoustic stimulation of the cochlea contralateral to the test cochlea (Contra Acoust); electrical stimulation at 140 pulses/s of the round window of the test cochlea (Ipsi RW 140 pps); electrical stimulation at 140 pulses/s of the OCB at the floor of the fourth ventricle (COCB 140 pps); or destruction of the cochlea contralateral to the test cochlea (Contra Damage). With electrical and acoustic stimulation, stimuli were applied for 1 minute. With contralateral cochlear damage, this cochlea was damaged about 2–5 minutes prior to the exposure.

from the IC contralateral to the test cochlea than from the ipsilateral IC. In fact, any particular amount of protection was obtained at a lower rate of stimulation at the contralateral IC than of the OCB in the brain stem or at the round window of the test cochlea (Figure 14-2, panel a).

5. Although higher centers (the IC) can activate the protective OCB system, the protection obtained by manipulations at the contralateral cochlea are exercised solely through lower brain stem pathways and do not require the intercession of any higher auditory cen-

ters. Thus, if the lower brain stem areas containing the pathways from the cochlea to the olivary complex and the return OCB pathways to the cochlea are isolated from all descending influences from higher auditory centers, manipulations at the contralateral cochlea can still protect the ipsilateral cochlea from NIHL.²²

6. The protection obtained with each of the various manipulations is graded to the NIHL that would otherwise ensue.¹⁹ This result was obtained in tests using a variety of 10 kHz exposures, with different exposure du-

rations and intensities. The effects are summarized in Figure 14-1 (panel c) where the amount of protection obtained at each frequency from 12 to 18 kHz in the test case (exposure combined with any one of the various protective manipulations) is plotted against the losses at each of these frequencies in the control groups (presenting only the monaural exposure). The line is that for an exponential function of the form $y = a((1 - \exp(-bx)))$, with the parameter a (the asymptote) set to 25 dB. The function appears to describe the data well, suggesting that the maximal amount of protection that could be obtained with OCB activation is about 25 dB.

7. All protective manipulations have persisting protective effects that outlast the application of the manipulation.^{2,8,9,19,20} Figure 14-2 (panel b) plots the amount of protection as a function of the delay between applying a test manipulation and presenting a standard loud sound exposure. In all cases except that of protection elicited by damage to the contralateral cochlea,^{5,19} significant protection was obtained even with a 5 minute delay between the 1 minute long manipulation and the subsequent loud sound exposure, although this protection was not as great as that obtained when the manipulation and exposure were applied together. With delays of 10 minutes or more, no protection was obtained. In the case of contralateral cochlear destruction, protection was more persistent, as might be expected from the nature of the manipulation that has no finite period of application, and significant, albeit decreasing, protection could be obtained even with 60 minute delay.

8. These persistent protective effects are not due to persistent effects at the cochlea, but are most likely to be exercised at the cell bodies of the protective OCB pathways.^{3-5,8,9,19} Thus, for example, the persistent effects of electrical stimulation of the OCB in the brain stem or of contralateral cochlear manipulations, can be blocked by lesioning the crossed OCB pathways in the brain stem in the period between applying the protective manipulation and the subsequent loud sound exposure.

9. Protective OCB-mediated effects are robust and resistant to deep anesthesia. The above-detailed experiments were carried out in animals deeply anesthetized with sodium pentobarbitone. Even without a muscle relaxant, in these animals it was generally difficult to obtain any evidence that other efferent pathways to the auditory periphery (i.e., those to the MEMs) were operative. (Similar results are detailed below from recent studies in the cat on OCB-mediated protection). Yet protection from loud sound exposures could be demonstrated robustly and reliably, using any of a variety of manipulations. Thus, protection could be elicited by direct activation of the OCB pathways by electrical stimulation in the brain stem,^{2,3,14} or the round window of the ipsilateral cochlea,^{9,10} by providing inputs from the contralateral cochlea either acoustically or electrically (R. Rajan, unpublished data, 1995), or by eliminating activity from the contralateral cochlea either by contralateral cochlear destruction^{5,8,16} or by the application of local anesthetics to the contralateral cochlea.¹⁶ In all cases it could be confirmed by specific pharmacological or surgical interventions that the protective effects were exercised through the (crossed) OCB pathways to the cochlea exposed to loud sounds. Thus, protective OCB effects are very robust and demonstrable by a wide variety of manipulations used to create the conditions leading to activation of the protective pathways.

10. In general, activation of the protective OCB pathways by manipulations at the contralateral cochlea or at the IC do not appear to directly activate the protective pathways. Instead it has been suggested^{4,5,8,19,20} that these manipulations provide facilitatory inputs to the cell bodies of the OCB neurons involved in protection, allowing the protective pathways to be more readily activated by the ipsilateral loud sound exposure.

Finally, in extrapolating from the above data, it is to be noted that protection could be obtained with contralateral acoustic stimulation with interaural intensity differences of up to 30 dB between the traumatizing ipsilateral sound and the contralateral protective low-

level sound. This result allows the protective effects to be more credibly translated into functional effects to be expected in the free field, and hence in real-life situations. Interaural intensity differences of up to 30 dB are recorded in a number of species at the higher frequencies (such as that used as the traumatizing exposure in these experiments) due to head-shadowing and pinna-amplification effects. If protection from traumatizing exposures could only be obtained when there were no interaural intensity differences (i.e., with the loud sound being equally loud in the two ears), it would be unrealistic to assume that this protection had any functional significance. Zero interaural intensity differences would only arise when a loud sound occurred at the midline, and therefore protection would arise only for loud sounds at this position. Because loud sounds can arise from any position in space, interaural intensity differences in the level of the loud sound would occur naturally. Even for sounds at the midline, interaural intensity differences would arise because of head and body movements that would change the relative levels of the sounds in the two ears. Thus, the fact that protection could be obtained even with large interaural intensity differences of up to 30 dB means that even if the loud sound were to be presented from one side of the head, resulting in traumatizing levels of sound in the ear on that side and the much lower (possibly nondamaging) levels on the other side of the head, protection of the traumatized ipsilateral ear could be obtained. From the guinea pig experiments it can be seen that the lower contralateral intensity would still be adequate to allow activation of the protective OCB pathways to the ipsilateral cochlea.

OCB-Mediated Protection from NIHL in Cats

In contrast to the above-detailed studies in guinea pigs, it has been reported^{17,18} that the OCB pathways did not provide any protection in cats exposed to loud sounds. The major study leading to this conclusion has been detailed previously¹⁸ and will not be repeated

here. In brief, in cats in which the cochleas on one side of the head were surgically deafened by a unilateral brain stem incision, binaural 1.5 or 6 kHz exposures produced similar threshold losses in the two ears. This was also the case when binaural 6 kHz exposures in unilaterally deafened cats were combined with brain stem electrical stimulation of the OCB pathways.

There are two major procedural differences between the cat study and the guinea pig studies: the frequency of the loud sound exposures, and the anesthetic used in the two species. Studies were carried out to determine if either or both of the factors could account for the difference in end outcome in efferent effects in the two sets of studies. Here, I present data from experiments using the same anesthetic agent as in the guinea pig studies to show that at least one of these factors adequately explains the differences between the two sets of studies, and that protective OCB-mediated effects can also be observed in the cat. These experiments also shed further light on other features of OCB-mediated protective effects.

In the experiments in guinea pigs, the animals were treated with a muscle relaxant to eliminate any possible confounding effects of the MEMs, which can attenuate loud sound transmission to the cochlea and thereby reduce threshold losses to loud sounds if activated. The drug used was shown to be very effective in blocking the activity of the MEM^{2,6} even when electrical stimuli were applied in the brain stem close to the course of the facial nerve pathways innervating one of these muscles.² In the experiments in cats reported here no relaxant was used. Instead experiments were first carried out to ensure that there were no confounding effects due to the MEM. Although a number of other studies have shown that the anesthetic agent (pentobarbitone sodium) used in these cat experiments severely depresses any activity of the MEM, it was decided to definitively establish whether the MEM played any role in the effects to be studied. To this end, in a number of animals the MEM in one ear were tenotomized using an RF cautery before presenting the ani-

mals with either sequential monaural exposures to each ear or simultaneous binaural exposures. Data for the latter condition are presented in Figure 14-3.

In the figure each panel presents data for the mean threshold losses recorded 5 minutes postexposure in groups of animals presented a binaural loud sound exposure at the frequency indicated in the top left-hand corner of each panel. In all animals the MEM were cut on one side but not on the other. The results for each group have been pooled as the losses recorded either in the ears with intact MEMs (MEM+) or in the ears in which the MEMs were cut (MEM-), over the appropriate affected frequency range. It is clear that the presence or absence of the MEM had no effect upon the threshold losses caused by binaural exposures at each of the three different exposure frequencies. There were never any significant differences between the mean threshold losses in the MEM+ ears compared to the MEM- ears. Thus, the MEM are not activated by these loud sounds in animals anesthetized with barbiturate, and have no effect on the threshold losses produced by binaural exposures at any of these three exposure frequencies.

Another series of experiments was then designed to test whether OCB-mediated protection could be obtained in the cat. In these experiments, a unilateral brain stem incision (of the type made by Liberman¹⁸ in his experiments in cats) was used to deafferent one cochlea. Then the animals were presented a binaural loud sound exposure to allow direct within animal comparisons of the effects of loud sound exposures on OCB-intact and OCB-cut cochleas. In different groups of animals different exposure frequencies were used. Threshold losses after the exposure were monitored in both ears in the same way as in all other animals in this study. Results for each of four different exposure frequencies are presented in Figure 14-4. (Other exposure frequencies were also used, but these four serve to illustrate the main point to be made here.) In the figure, each panel presents data for the mean threshold losses recorded 5 minutes postexposure after binaural exposure at the

exposure frequency indicated in the top left corner of each panel. Data for each exposure frequency (i.e., for each group) have been grouped into the losses occurring in the ears to which the cochlear efferents were cut in the brain stem (OCB- ears in figure legend) or the losses occurring in the ears with intact cochlear efferents (OCB+ ears in figure legend).

The figure illustrates a clear frequency difference in the protective effect of the cochlear efferents. In the case of binaural 3 or 7 kHz exposures, there were no differences between threshold losses in the OCB+ ears and those in the OCB- ears, that is, there were no protective OCB effects for these exposure frequencies. (These frequencies are very similar to those used by Liberman.¹⁸) In contrast, there were very large and significant differences between OCB+ and OCB- ears in the case of 11 or 20 kHz exposures. In both these groups, OCB+ ears suffered significantly less threshold losses than did OCB- ears. Thus, for these exposure frequencies, there were very significant OCB-mediated protective effects. (Note that the frequency of 11 kHz is almost identical to that used in the guinea pig studies.)

It is worthwhile noting that Figure 14-3 demonstrates that the MEMs were not operative with binaural exposures at 11 kHz, while the frequency of 20 kHz, at which strong protective effects were also observed, is known to be well outside the range of frequencies that activates or is affected by contraction of the MEM.^{23,24} In this context it is also to be noted that the direction of the frequency dependency of OCB-mediated protective effects is opposite to that expected from effects due to the MEM. The latter are most effectively activated at low frequencies and have their greatest effects at these frequencies. The converse is true here with the protection demonstrated in these experiments. Thus, in addition to the effects in Figure 14-3, the pattern of frequency dependency also shows that the MEM are not involved in the protective effects.

In addition to demonstrating that the OCB pathways do exercise protective effects in the cat (as in the guinea pig), these data also suggest that the protective effects of the OCB only

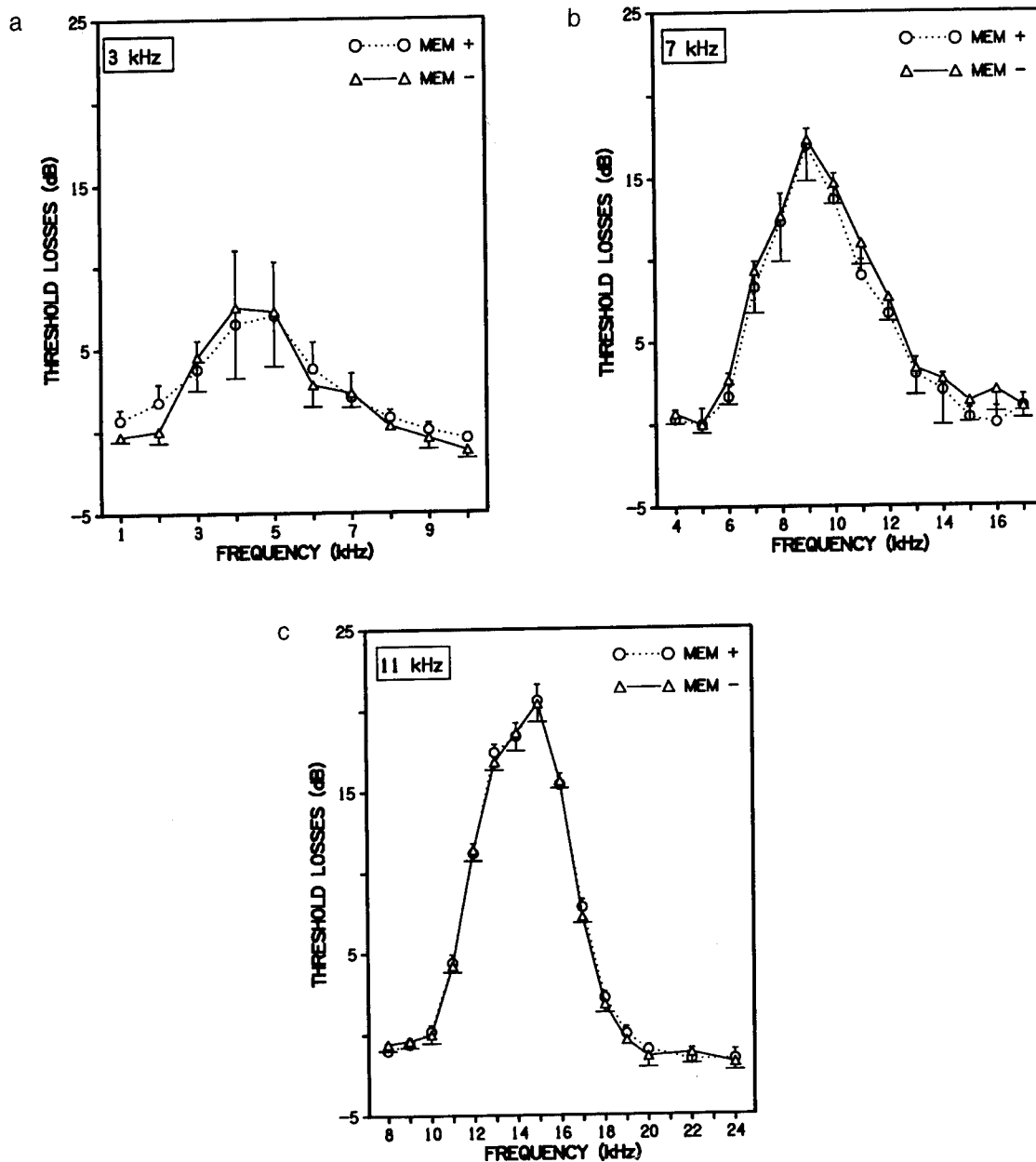


Figure 14-3 Absence of middle ear muscle (MEM) effects in binaural loud sound exposures in barbiturate-anesthetized cats. Each panel presents the mean threshold losses (error bars = SEM) 5 minutes postexposure, in groups of animals presented binaural loud sound exposures at 3 kHz (a), 7 kHz (b), or 11 kHz (c). Exposures were at 100 dB SPL for 10 minutes. In each animal the MEMs in only one ear were cut with an RF cautery. The data for each exposure frequency have been grouped as the losses occurring in the ears with intact MEMs (MEM+) or the losses in the ears in which the MEMs were cut (MEM-).

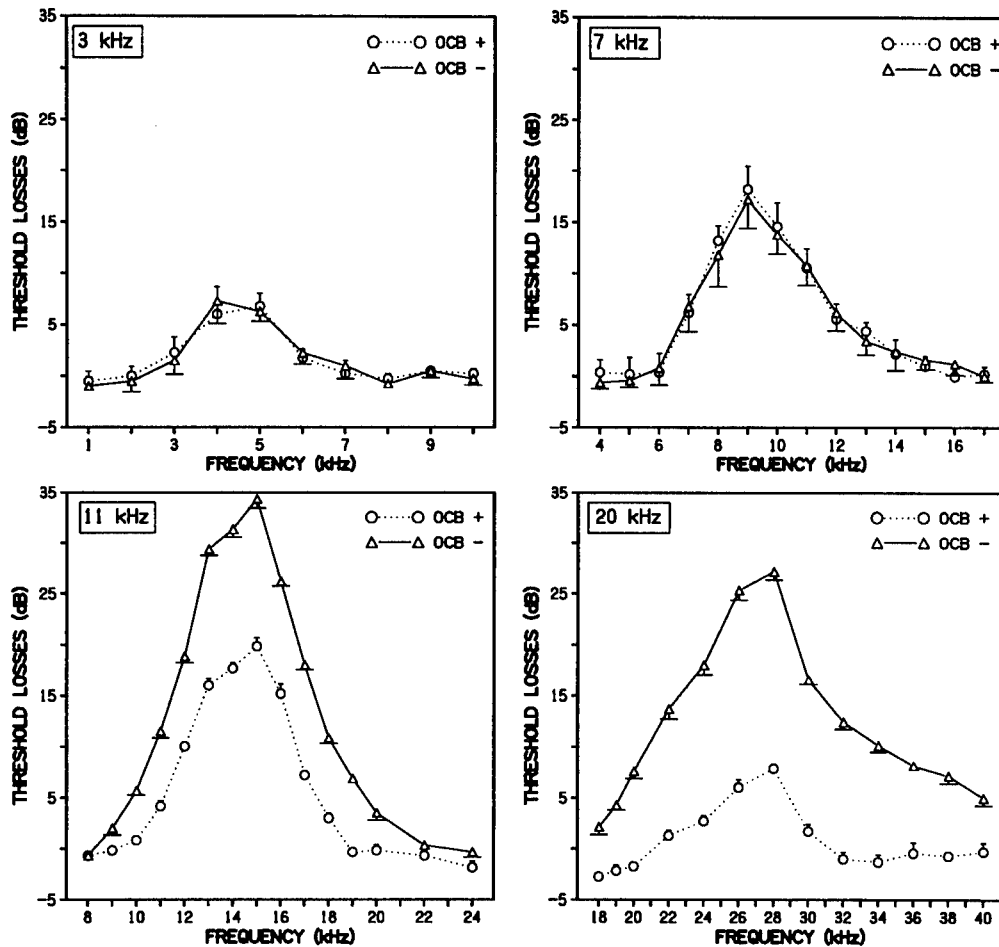


Figure 14-4 Protective OCB effects in cats at different exposure frequencies. Each panel presents the mean threshold losses (error bars = SEM) 5 minutes postexposure, in groups of cats presented binaural loud sound exposures at 3 kHz (top left), 7 kHz (top right), 11 kHz (bottom left), or 20 kHz (bottom right). Exposures were at 100 dB SPL for 10 minutes. In all cats a unilateral brain stem incision was made along the floor of the fourth ventricle at a point about 2 mm away from the midline to cut the OCB only to the cochlea on that side. Data for each exposure frequency have been grouped as the losses in ears with intact OCB (OCB+) or those in deafferented ears (OCB-).

occur for high-frequency exposures but not for low-frequency exposures. However, from other studies in this series, it appears that the difference may not be specifically related to the exposure frequency alone but also to the NIHL produced. In the guinea pig studies, as noted above, there was a very strong dependency of the amount of protection on the losses that would otherwise ensue. Similar effects also occur for OCB-mediated protection in the cat (not illustrated here). As can be seen from Figure 14-4, there is a very clear

difference in the NIHL caused by the different exposure frequencies, even though they were all at the same intensity and for the same duration. This can be appreciated in examining the losses recorded in the four groups in the OCB- ears in which protective OCB effects (for the high-frequency exposures) do not act as a confounding factor in this comparison. The traumatizing effects of the different frequencies can be ordered in terms of the peak losses in cochlear sensitivity produced, resulting in an ordering (from greatest to least dam-

age) of 11, 20, 7, and 3 kHz exposures. Similar differences in susceptibility to NIHL, over a more limited range of exposure frequencies (2, 4, and 8 kHz) were found by Decory and colleagues²⁵ in each of three species (cats, guinea pigs, and chinchillas). In the latter study it was found that the order of traumatizing exposures (from most to least traumatizing) was 8, 4, and then 2 kHz. This ordering is in the same direction as the ordering on the basis of NIHL of the three lowest exposure frequencies in the present study (11, 7, and 3 kHz), with greatest NIHL being produced by the 11 kHz exposure and least NIHL by the 3 kHz exposure. Thus, over this range of frequencies, at least, the lower the exposure frequency the less the NIHL produced.

Given these two sets of effects, it is therefore possible that OCB-mediated protection can be found for lower frequencies of exposure provided that the exposures are sufficiently traumatizing. In other studies I have found that this is at least partly true, although not necessarily for the lowest exposure frequencies. Thus, there may not necessarily be a frequency dependency to OCB-mediated protection except for the lowest exposure frequencies. It is interesting to note that the low frequencies are those that are most efficient at activating the MEM and the ones most affected by contraction of the MEM. Thus, if MEMs also have a protective role, it is possible that the two systems may not overlap in the regions of the cochlea subject to these protective effects.

In general these data affirm that the OCB can also act to protect the cochlea in the cat. Recent studies have also found this OCB-mediated protection in the rat,¹⁵ confirming the generality of this functional role across species. It is to be emphasized here, as noted in my previous reports and reviews, that there is very good (albeit circumstantial) evidence that the protective effects are exercised only by one subcomponent of the OCB. Detailed studies in the guinea pig suggest that this component is most likely the crossed OCB pathways emanating from ventrally located nuclei in the superior olivary complex, and

mediating effects on the cochlea's outer hair cells via nicotinic cholinergic synapses. There is currently no evidence that addresses the issue of whether it is the same component of the OCB pathways in the cat that acts in the protective mode. Given the general nature of the effects and the similarity of the effects to those seen in guinea pigs, it is most parsimonious to assume that it is the same component of the OCB pathways that exercises the protective effects in cats.

Interaction Between OCB-Mediated Protection and Priming Protection from NIHL

In an earlier study²⁶ it was shown that the NIHL caused by a loud sound exposure could be modulated by "priming" a cochlea with sound prior to the test exposure. Here, an initial monaural low-level sound exposure could significantly reduce the NIHL caused by a subsequent high-level loud sound exposure to the same cochlea, even when there was no residual NIHL to the priming sound. Thus, the reduction in threshold losses to the high-level test exposure was not due to the presence of residual losses to the initial priming low-level exposure. Studies of this priming effect have since been extended by others into more detailed studies of "toughening" effects in NIHL,²⁷ and other chapters in this book address the features of such toughening effects. We have examined whether the two sets of effects modulating NIHL were independent by testing whether the priming effects were mediated by OCB pathways. Results from these studies are presented here. The experiments were carried out under similar conditions to those used in guinea pigs in the studies on OCB-mediated protective effects: the animals were anesthetized with Nembutal and were paralyzed with a muscle relaxant to eliminate the influence of the MEMs. Here all tests were carried out only monaurally.

The basic priming effect on the losses caused by the high-level exposure are shown in panel **a** of Figure 14-5. In the control group

the high-level exposure was presented by itself; in the test group a priming low-level monaural exposure had been presented to the same cochlea about 40 minutes previously. The priming exposure produced small threshold losses (maximally less than 20 dB at the half-octave point 10 seconds after the priming exposure) and threshold sensitivity recovered within about 15 minutes of the priming sound. Thus, by the time the high-level exposure was presented in the test group, about 40 minutes after the priming sound, there were no residual losses left from the priming exposure, and cochlear sensitivity was similar to that at the start of the experiment. Despite this, significantly lower threshold losses were recorded over the most affected frequency range (from about 10 to 20 kHz) after the high-level exposure in the primed test group compared to the losses recorded in the same exposure in the control group (Figure 14-5, panel a).

In two further groups of animals, tests were carried out to determine whether this protective effect was mediated by the same OCB pathways as were responsible for the protection detailed above. This was ascertained by testing with a pharmacological agent (at a much higher dose) that was effective in blocking the OCB-mediated protection detailed above. In one of the two new test groups, an IP injection of strychnine at 10 mg/kg body weight was administered about 15 minutes prior to the priming sound. In the other group the same drug at the same dose was administered about 15 minutes after the end of the priming sound. Thus, effectively, in one group the strychnine was administered about 25 minutes prior to the high-level exposure and in the other group about 55 minutes prior to the high-level exposure.

It is to be noted that this drug at a much lower dose was effective in totally blocking the protective efferent effects detailed above. In those studies, this drug at a lower dose of 4 mg/kg body weight blocked the protective effects of contralateral cochlear manipulations within 15 minutes of IP administration, and blocked the protective effects of direct electrical activation of the efferents within 45–60 minutes.

(The difference in the time course in these two cases has been discussed previously, see Rajan,¹⁹ and will not be detailed here.)

The results from these two groups of animals are shown in panel b of Figure 14-5, where they are compared to the losses recorded in the two previous groups. As shown in the figure, strychnine treatment did not, in either of the two new test groups, affect the fact that the priming sound still was able to reduce the damage caused by the subsequent high-level exposure. (Note, as detailed elsewhere^{2,8} that strychnine did not alter the losses caused by the monaural priming exposure itself, or in any way modify the time course of recovery from the small threshold losses caused by the priming sound.) In both these two primed test groups, the high-level exposure resulted in smaller threshold losses than in the case of the control group. The threshold losses in these two groups were not significantly different from the losses recorded in the previous test group with priming but without any strychnine being used.

Thus, these results show that the priming protection is not mediated by the same OCB system as is responsible for the protection detailed above. It is also to be noted that these experiments were carried out in animals treated with a muscle relaxant to eliminate the influence of the MEMs, using a drug shown to be effective in doing so.^{2,6} Thus, the priming protection is also not mediated by the MEM.

Although these data exclude the possibility that the same OCB system mediates protection in both sets of NIHL modulations, it does not mean that the OCB is not involved in priming protection. As emphasized above, the OCB-mediated protection discussed is likely to be mediated only by one of the subsystems of the olivocochlear efferent pathways to the cochlea. Thus, it is possible that other subsystems of the OCB may be responsible for the priming protection. This possibility needs to be specifically tested before any OCB role in priming protection can be excluded. On the basis of this evidence it can only be stated that the crossed OCB nicotinic system that has been implicated in the protec-

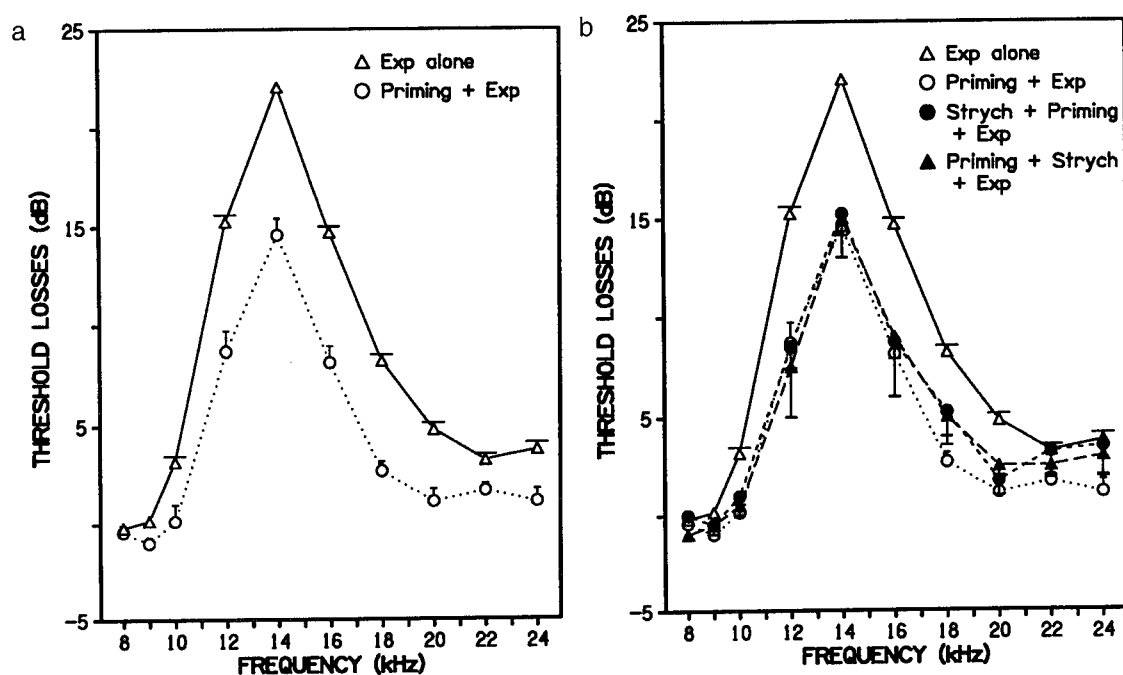


Figure 14-5 Priming modulation of NIHL in guinea pigs. Each panel presents the mean threshold losses (error bars = SEM) recorded 5 minutes after a standard monaural loud sound exposure (always at 10 kHz, 103 dB SPL for 1 minute) in each group of animals. Panel **a** shows the basic priming effect and panel **b** shows data from tests conducted to determine if the OCB was involved in the priming effects. The priming sound was an exposure at 10 kHz, 97 dB SPL for 1 minute. When used, this priming sound was presented about 40 minutes before the standard exposure. (**a**) Control group (Exp alone): only standard exposure presented. Test group (Priming + Exp): animals were first presented an exposure at 10 kHz, 97 dB SPL for 1 minute. About 40 minutes after this priming exposure the animals were presented the standard exposure. Losses are those after the standard exposure. (**b**) Data from the two groups in panel A (Exp alone, and Priming + Exp groups) are presented along with data for two new groups tested with strychnine injected at an IP dose of 10 mg/kg body weight. In one group, strychnine was injected about 15 minutes prior to the priming sound (Strych + Priming + Exp). In the other group the strychnine was injected about 15 minutes after the end of the priming sound (Priming + Strych + Exp). Effectively, in these two groups, strychnine was injected either about 25 minutes prior to the standard exposure (Priming + Strych + Exp) or about 55 minutes prior to the standard exposure (Strych + Priming + Exp).

tion from NIHL obtained with contralateral manipulations, and with stimulation at higher auditory centers, is not the system involved in the monaurally derived priming protection.

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Chapter 15

Protection Against Temporary and Permanent Noise-Induced Hearing Loss by Sound Conditioning

Barbara Canlon and Safak Dagli

Protection Against Noise Trauma

The mechanical energy of sound can be detrimental to the mammalian hearing organ, the cochlea. The detrimental effects on the cochlea can be either temporary or permanent in nature. At present, the underlying mechanisms that distinguish a temporary threshold shift from a permanent threshold shift are not yet understood. One means of describing the overall insult of acoustic energy and its relation to the anatomy and physiology of the cochlea is the equal energy principal. The equal energy principal, first proposed by Eldred in 1955¹ states that the degree of noise-induced hearing loss is directly related to the total acoustic energy. A reciprocal relationship between the intensity and duration of the exposure allows for any type exposure to be equally equated for total energy. Accordingly, noise exposures of equal energy should yield similar degrees of hearing loss. For many different types of continuous exposures, the equal energy principal is a close approximation to the subsequent hearing loss.² However, there are many instances where the equal energy principal is clearly not a good indicator for the resultant hearing loss.³ The suitability of the equal energy hypothesis and the subsequent detrimental effects of acoustic overstimulation on the cochlea is questioned immediately when one considers that there are a variety of methods that can be used to alter the sensitivity of the cochlea to acoustic

trauma. For example, manipulation of cochlear metabolism is one means of reducing or augmenting the damage induced by noise. Increasing or decreasing body temperature during noise exposure has been shown to increase or decrease the consequent damage.^{4,5} In addition, increasing the oxygen supply or removing the thyroid gland can also protect the ear from noise-induced hearing loss.⁶ Recently, protection against noise-induced hearing loss was shown by the inhibition of free oxygen radical induced lipid peroxidation.⁷ As a general rule, it seems as if the metabolic state of the cochlea plays an important role in determining the degree of noise-induced hearing loss. An increase in high energy reserves may afford a resistance to metabolic exhaustion.

Activation of the medial cochlear efferents has also been shown to modify cochlear sensitivity to noise trauma in the guinea pig. It has been shown that protection of the ipsilateral ear occurred when the contralateral ear was simultaneously stimulated at the same frequency but at a lower intensity.⁸ This effect was blocked by strychnine, a known blocker of the medial efferent system of the inner ear. Furthermore, high rates of electrical stimulation of the crossed olivocochlear bundle presented simultaneously with acoustic stimulation reduced the magnitude of a temporary threshold shift.⁹

Another means of experimentally reducing the detrimental effects of noise trauma is by

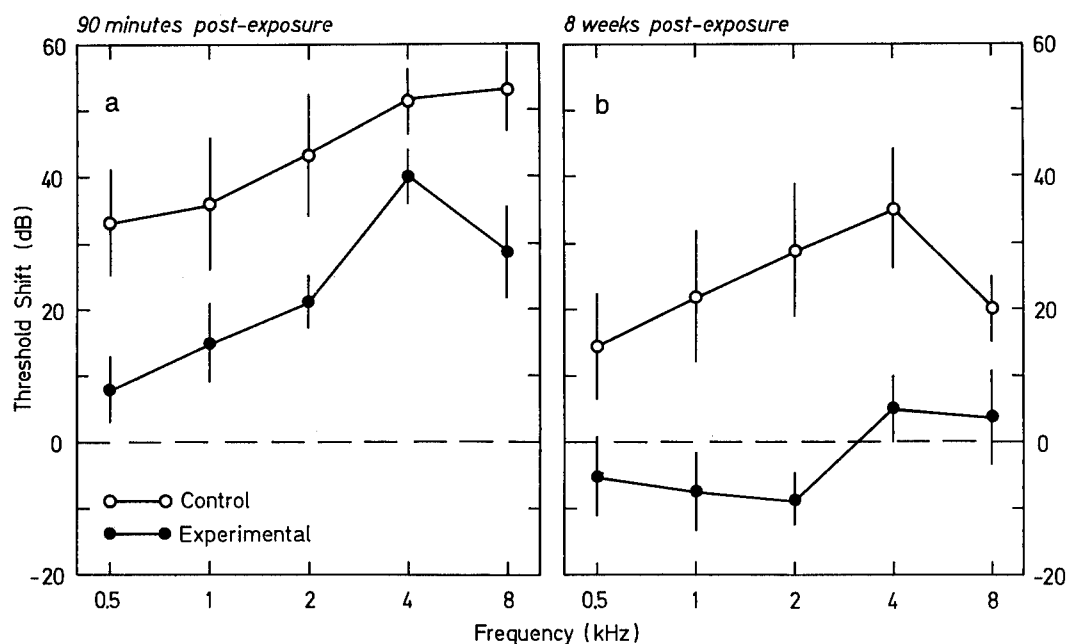


Figure 15-1 Auditory brain stem threshold shift induced by a traumatic noise exposure (1 kHz, 105 dB SPL, 72 hours) measured either (a) 90 minutes or (b) 8 weeks postexposure. (○) Control, unconditioned group and (●) sound conditioned group.

sound conditioning. Protection against noise trauma can occur when guinea pigs are sound conditioned to a continuous low-level, long-term, nondamaging stimulus before the traumatizing exposure.^{10,11} Sound conditioning provides 20 dB protection against a traumatizing stimulus compared to a control, unconditioned group. After either 1 or 2 months of recovery, the sound conditioned group showed complete recovery, but the control group continued to show a threshold shift between 20 and 30 dB (Figure 15-1). The continuous sound conditioning paradigm has been proven effective for protecting against the permanently damaging effects of noise trauma in the guinea pig,^{10,11} rabbit,¹² and gerbil.¹³

Another way of providing protection against noise trauma is by using interrupted repetitive stimulation. When the interrupted repetitive paradigm is employed, a threshold shift is induced during the initial days of the exposure and gradually recovers during the remaining days of exposure.¹⁴⁻¹⁷ In addition, the interrupted repetitive paradigm has been shown to afford protection against a sub-

sequent traumatizing exposure in the chinchilla,^{18,19} rabbit,²⁰ guinea pig,²¹ and of special interest, human subjects.²²

The main aim of this review is to describe the effects of continuous sound conditioning on a subsequent temporary, as well as a permanent noise-induced hearing loss using both morphological and physiological techniques.

Distortion Product Otoacoustic Emissions and Effect of Permanent Hearing Loss

When stimulating the cochlea with two primary tones, f_1 and f_2 , the auditory system responds in a nonlinear fashion by generating distortion products. This distortion is converted into acoustic energy that can be recorded with a sensitive microphone placed in the ear canal. The most readily detectable distortion product is the lower cubic difference tone ($2f_1 - f_2$). When recorded in the ear canal of guinea pigs, the level of the cubic difference tone typically lies 20–40 dB below the level of the primaries. The amplitudes of the distort-

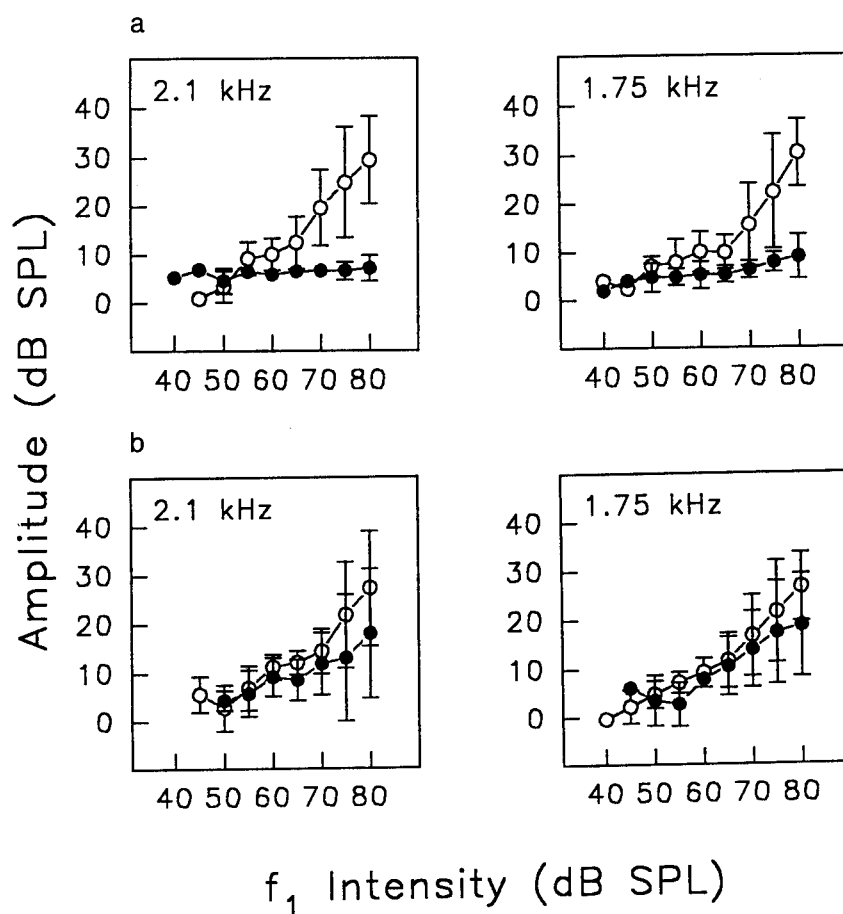


Figure 15-2 Input–output function of the distortion product otoacoustic emission for the (a) control, unconditioned group and (b) the sound conditioned group. Values are the mean and standard deviation for measurements made (○) before and (●) 4 weeks postexposure (1 kHz, 105 dB SPL, 72 hours).

tion products are dependent on the frequency relation of the primary tones, but also reflects cochlear status at the region where the primaries interact.

The most prominent distortion product emission is the cubic distortion product $2f_1 - f_2$. There is a 30–40 dB increase in the amplitude of the emission with an increase of 40 dB f_1 stimulus level, corresponding to a 1 dB or slightly below 1 dB increase for each 1 dB increase in the primary stimulus. At approximately 70 dB sound pressure level (SPL) there often was a decrease by 5–10 dB, after which the emission grew again with increasing stimulation.

The mean and standard deviation values for the distortion product amplitude as a function of f_1 intensity are illustrated in Figure 15-2a (upper panel) for the control group and for the sound conditioned group (lower panel). There are two different frequencies (2.1 and 1.75 kHz) represented in this figure with preexposure (open circles) and 4 week posttraumatizing exposure (closed circles) represented. The control preexposure distortion product amplitude values at frequencies 2.1 and 1.75 kHz, show approximately a 30 dB increase over a 40 dB f_1 intensity range. When studied 4 weeks after the traumatizing tone, the distortion product amplitudes are reduced for all

frequencies. There is no growth of the distortion product emission at 2.1 and 1.75 kHz, despite an increase in f_1 over a 40 dB range.

The sound conditioned group demonstrates similar preexposure distortion product amplitudes as the control group (Figure 15-2b, lower panel). There is a 30 dB increase in distortion product amplitude as the intensity of f_1 increases over a 40 dB range. The effect of the traumatic noise on the distortion product amplitudes from the sound conditioned group are illustrated by the filled circles in Figure 15-2. For the 2.1 and 1.75 kHz frequencies the amplitude of the distortion product increases by approximately 15–20 dB as f_1 intensity increases. At higher f_1 intensities, the postexposure values are lower than the preexposure values between 5 and 10 dB.

For statistical analysis the input–output distortion product emission curves were condensed to one single value by taking the area under the curve. The area under each curve was calculated for each animal at each f_1 stimulus intensity both before exposure to the 105 dB SPL tone and 4 weeks posttrauma. Comparisons were made for the control group before and 4 weeks posttrauma, as well as for the sound conditioned group before and 4 weeks posttrauma. The control group show statistically significant (t test, $p < 0.05$) differences after the 105 dB traumatizing stimulus. This difference indicates that the area under the input–output distortion product emission curve does not grow with increasing stimulation intensity as does the preexposure values. The sound conditioned group, on the other hand, does not show statistically significant differences between preexposure and postexposure values.

Continuous Sound Conditioning Protects Outer Hair Cells

The threefold objective of analyzing surface preparations of the organ of Corti from guinea pigs was to: identify the hair cell type affected by the traumatic noise; quantify the degree of hair cell loss; and determine if the pattern of loss was different in the sound conditioned

group compared to the unconditioned group.

After a recovery period of 1 month from the traumatic exposure (1 kHz, 105 dB SPL, 72 hours) cochleae were removed from both the control, the unconditioned group, and the conditioned group. Surface preparations were stained with fluorescently labeled phalloidin (Molecular Probes, USA). Phalloidin was used to label structures containing filamentous actin. Because the stereocilia and the cuticular plate are actin bearing structures, they react intensely with phalloidin. The use of phalloidin has eased the burden of counting the hair cells as well as determining stereocilia and cuticular pathology. After counting the missing hair cells a cochleogram was constructed showing the percent hair cell loss and the distance from the round window. An approximate frequency map related to the distance from the round window is indicated. The length of the guinea pig cochlea is estimated to be 18 mm. According to Békésy,²³ there is approximately a 2–2.5 mm distance between each octave. It is estimated that the 9 mm distance from the round window represents the 8 kHz region; the 11 mm region represents the 4 kHz region; the 13 mm region represents the 2 kHz region; and the 15 mm distance represents the 1 kHz region.

The control group, exposed to the traumatic stimulus only, showed a significant degree of outer hair cell (OHC) loss in the middle region of the cochlea (Figure 15-3a). There is nearly 100% loss of all three rows of OHCs in the 12–14 mm distance from the round window. Slight individual variations are evident and could partly be due to small irregularities in the dissection of each cochlea, or due to the individual guinea pigs susceptibility to noise trauma. The inner hair cells (IHCs) in the control animals were intact in nearly all the animals. In most of the animals the three rows of OHCs were affected to the same degree. There was no apparent gradation amongst the three rows of OHCs with respect to their sensitivity to trauma. The general pattern of loss in the control animals is such that the region of maximal loss is located within one particular area on the basilar membrane. This area usu-

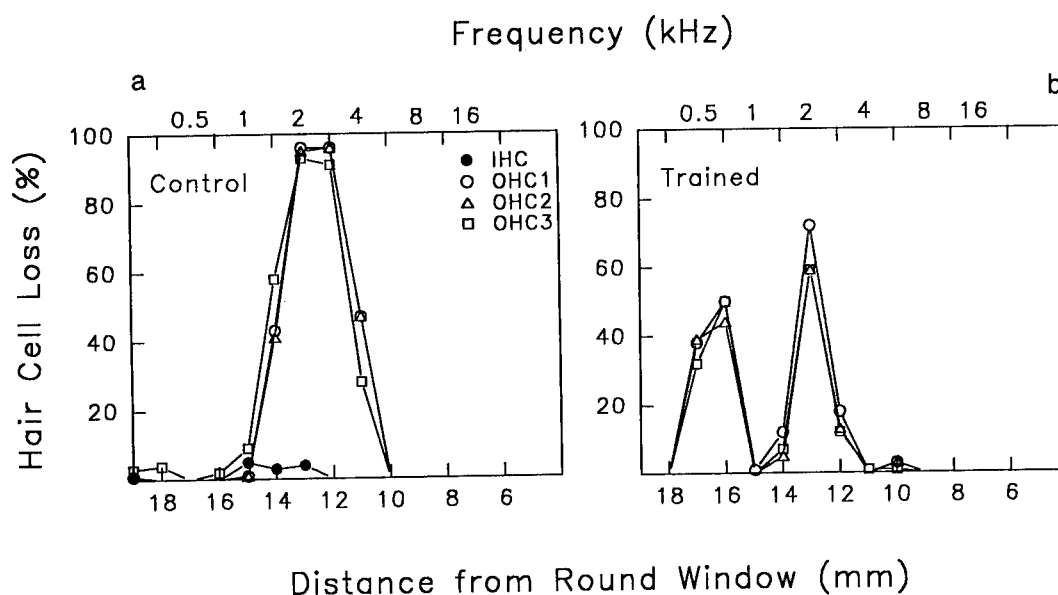


Figure 15-3 Representative cochleograms 4 weeks after the traumatic exposure for (a) a control and (b) a sound conditioned animal. The percent hair cell loss versus the distance from the round window is represented with an approximate frequency scale.

ally extends over a 3–4 mm distance in the region corresponding to 1–4 kHz.

The sound conditioned group showed a strikingly different pattern of hair cell loss compared to the control group (Figure 15-3b). Instead of having a maximal loss located within one particular area on the basilar membrane as the control group, the sound conditioned group usually illustrated two distinct regions of damage, the first region being approximately at 16 mm distance and the other at approximately 13 mm distance from the round window. All three rows of OHCs are affected equally and there is no indication of a graded damage among the different rows of OHCs. The IHCs are not affected by the traumatic exposure.

It is also evident from Figure 15-3 that the amount of OHC loss in the sound conditioned group is much less compared to the control, unconditioned group. The sound conditioned animals show between a 50 and 70% loss of OHCs.

It is interesting to note that the pattern of OHC loss induced by the traumatic noise ex-

posure in the sound conditioned animals is altered. While the cause for this localized protection of OHCs is unknown, the findings may indicate that this region acts as a center point in the protection phenomenon. The regions flanking this center point are also protected as revealed by the lower degree of loss. Previous findings on cochleae from sound conditioned animals show a reduced calbindin immunoreactivity, a calcium binding protein with a potent ability to buffer intracellular calcium.²⁴ A decreased calbindin immunoreactivity may inhibit calcium influx into the cell during prolonged periods of continuous stimulation. A consequence of the reduced calcium influx may make the cell less excitable and thereby protect the cell from overstimulation. During the kindling model of epilepsy, calbindin shows a progressive decline in the dentate granule cells in the hippocampus²⁵ and these authors have proposed that a reduction in intracellular calbindin would reduce the buffering capacity of the cell and cause a more rapid inactivation of calcium channels making the cell less excitable.

Continuous Sound Conditioning Influences Magnitude and Rate of Recovery from Temporary Threshold Shift

The amplitudes of the distortion product otoacoustic emission ($2f_1 - f_2$) were examined systematically in two groups of guinea pigs, a sound conditioned (1 kHz, 81 dB SPL, 24 days) and an unconditioned group. The amplitude of the distortion product was followed after daily exposures to a 2767 Hz tone at 103 dB SPL for 5 minutes. This exposure was deter-

mined to cause a temporary threshold shift in control animals. For a period of 3 consecutive days the exposure was delivered in a closed system and the distortion product emission followed at 60 dB SPL at the following frequencies: 4.4, 3.5, 2.8, and 2.2 kHz. Each day before overstimulation the distortion product amplitudes were monitored for intensities between 35 and 80 dB SPL.

The effect of 3 days of overstimulation on the control animals is shown in Figure 15-4. The amplitude of the distortion product otoacoustic emission prior to overstimulation is

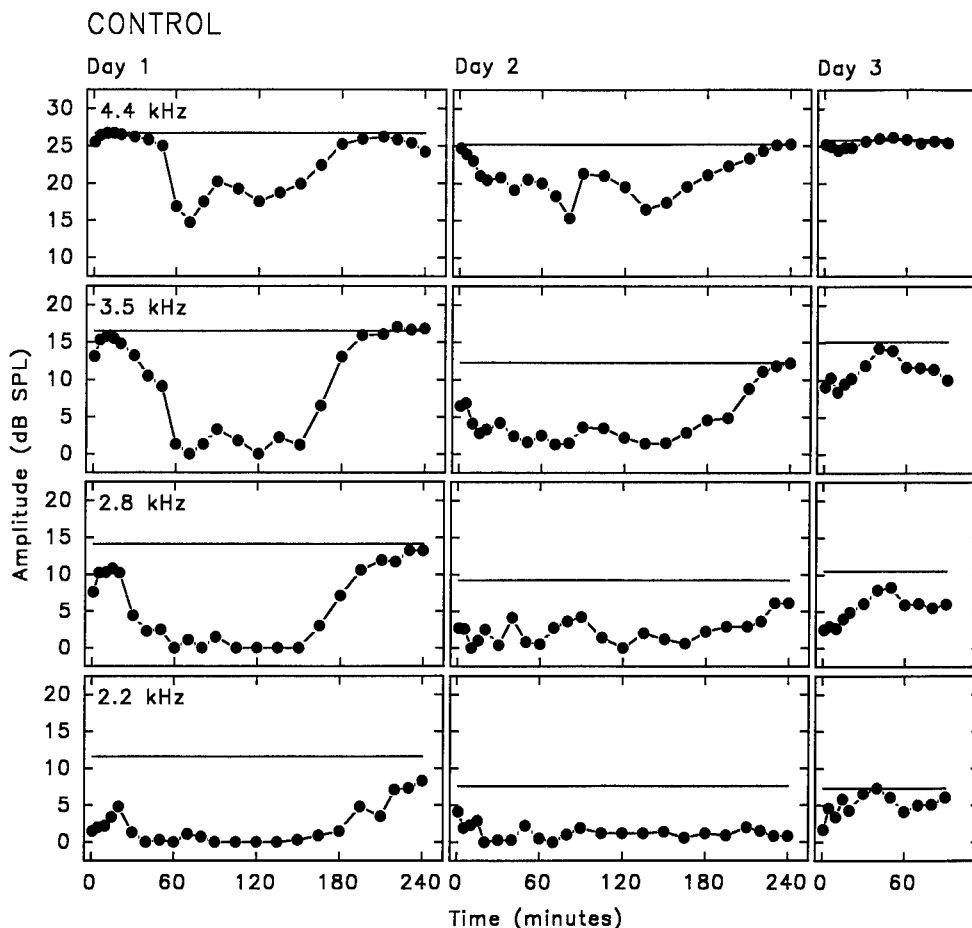


Figure 15-4 Acute effects of a temporary threshold shift on the amplitude of the distortion product otoacoustic emission from a control, unconditioned animal. (—) The amplitude of the distortion product emission was recorded prior to each exposure. (●) The change in the amplitude of the distortion product emission was followed for 240 minutes the first 2 days and 90 minutes on the third day. Exposures were repeated daily for 3 days.

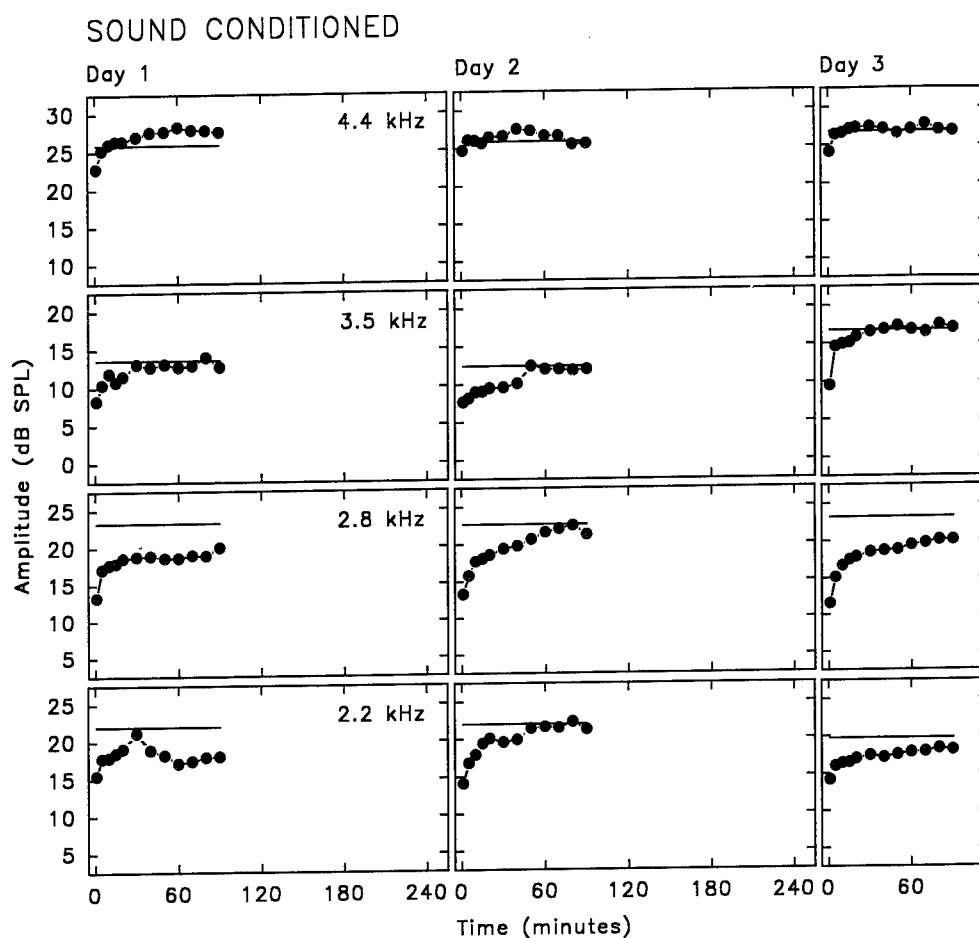


Figure 15-5 Acute effects of a temporary threshold shift on the amplitude of the distortion product otoacoustic emission from a sound conditioned animal. The amplitude of the distortion product emission was recorded (—) prior to each exposure and (●) for 90 minutes postexposure. Exposures were repeated daily for 3 days.

represented by the solid line. Directly after the overstimulation only minor alterations are noted for the amplitude of the distortion product. An interesting feature of these curves is the delayed response of the insult. It is not until 60 minutes postexposure that the amplitude declines to the noise floor at 3.5, 2.8, and 2.2 kHz. Between 60 and 180 minutes postoverstimulation the amplitude is maximally affected by 10–15 dB. This is particularly evident for 3.5, 2.8, and 2.2 kHz. The cause for this delay is not known, but is a consistent finding among the different control animals. Because the delay is as long as 60 minutes, suggests that metabolic changes are occur-

ring. On the first 2 days of overstimulation the control animals required 240 minutes for recovery, the third day of overstimulation required only 90 minutes. The preoverstimulation values obtained on day 2 were often depressed by approximately 5 dB, despite near normal distortion product amplitudes at the end of day 1 for 3.5, 2.8, and 2.2 kHz. In general, with repeated stimulation the amplitude of the distortion product emission becomes less affected by the overstimulation.

In contrast to the control group, the animals preexposed to the sound conditioning paradigm were significantly less affected by the overstimulation at all tested frequencies (Fig-

ure 15-5). The difference was apparent in both the magnitude of the depression as well as the rate of recovery. In the sound conditioned group, the 4.4 kHz distortion product emission is nearly always unaltered by the overstimulation. The other frequencies show between a 5 and 10 dB shift that occurs immediately postexposure. It was always feasible to record the amplitude of the distortion product emission from the sound conditioned group because their values were above the noise floor. Complete recovery from overstimulation required only 60–90 minutes compared to the near 240 minutes in the control animals. In fact, most of the recovery was achieved within the first 30 minutes. These results demonstrate that the sound conditioned animals are affected by a temporary threshold shift, yet have the capacity to recover more rapidly, most likely due to the small initial shift induced by the overstimulation.

Conclusions

Sound conditioning protects the peripheral auditory system from a subsequent noise trauma. When compared to a control group the consequence of a subsequent traumatic exposure: reduces the degree of outer hair cell loss; causes an altered pattern of damage; maintains the amplitude of the distortion product ototacoustic emission over a wide frequency range for a permanent noise-induced threshold shift; and increases the rate of recovery and reduces the magnitude of a temporary noise-induced threshold shift.

The results described in this summary indicate that the degree of hearing loss is reduced by exposure to a low-level, long-term, non-damaging exposure prior to a traumatizing exposure of either a temporary or permanent nature. The anatomical site responsible for the protection against noise trauma is not yet known. Evidence is accumulating that supports the notion that the intrinsic properties of the OHCs are modified by sound conditioning. The findings presented here provide a foundation on which to further assess and experimentally test the differences between tem-

porary and permanent noise-induced hearing loss.

Acknowledgments

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Chapter 16

Psychophysical and Evoked Response Studies of Aged Subjects: Masking by Low-Pass Noise

John H. Mills, Flint A. Boettcher, Judy R. Dubno,
and Richard A. Schmiedt

Persons with sensorineural hearing loss commonly report difficulty hearing and recognizing speech and other auditory signals in the presence of a competing noise. The experimental study of this ubiquitous phenomenon has produced many studies of the masking of speech and other signals by maskers that vary in level and spectrum. Several quantitative methods have evolved that attempt to predict the performance of an individual or groups of individuals, especially the hearing impaired, on tests of speech perception. Often these tests are conducted under a wide array of experimental conditions including masker spectrum, signal-to-noise ratios, types of speech material, filtering of the signal, reverberation times, and so on. Of all of the various schemes that have evolved, the Speech Transmission Index and the Articulation Index (AI) still enjoy considerable popularity.

As part of a longitudinal, large-scale study of age-related hearing loss, we measured a large number of auditory behaviors in a sample of about 200 older persons. Included in this auditory test battery were several measures of speech perception in quiet and in noise. These and other measures were used in conjunction with the AI to predict speech recognition of a large group of older persons in quiet (NU-6 words) and in babble (SPIN sentences). Results are given in Figure 16-1.¹ Note that in the top panel, which is the quiet listening condition, the use of audiometric thresholds and the AI to predict NU-6 results was reasonably successful. About 85% of the ob-

served scores were accurate within $\pm 10\%$ of the predicted scores. Results in babble (lower panel) are less satisfactory. Difference scores exceed 20% for a substantial portion ($>60\%$) of the sample. Here, we do not attempt to resolve the shortcomings of the AI in predicting SPIN results in babble. Rather, we use this example to show that despite some 40–50 years of research directly or indirectly targeted at predicting speech intelligibility in noise, errors of greater than 20–25% occur in more than 60% of the sample.

One factor that almost surely plays a major role in the perception of speech in noisy circumstances is upward spread of masking. This masking effect refers to the masking of a high-frequency signal by a low-frequency masker and was first described by Wegel and Lane.² Since 1924 there have been many studies directly or indirectly aimed at the anatomical/physiological bases of upward spread of masking, or in different terms, the asymmetrical masking pattern produced by a low-frequency masker. Clearly, part of this asymmetrical masking pattern reflects the Békésy travelling wave as it proceeds from the base to the apex of the cochlea. What other mechanisms are involved? What is the influence of hearing loss? Is upward spread of masking explainable by unusually poor tip-to-tail ratios of tuning curves or a loss of two-tone rate suppression?

Perhaps the most impressive example of upward spread of masking was reported in 1970 by Martin and Pickett.³ In three subjects

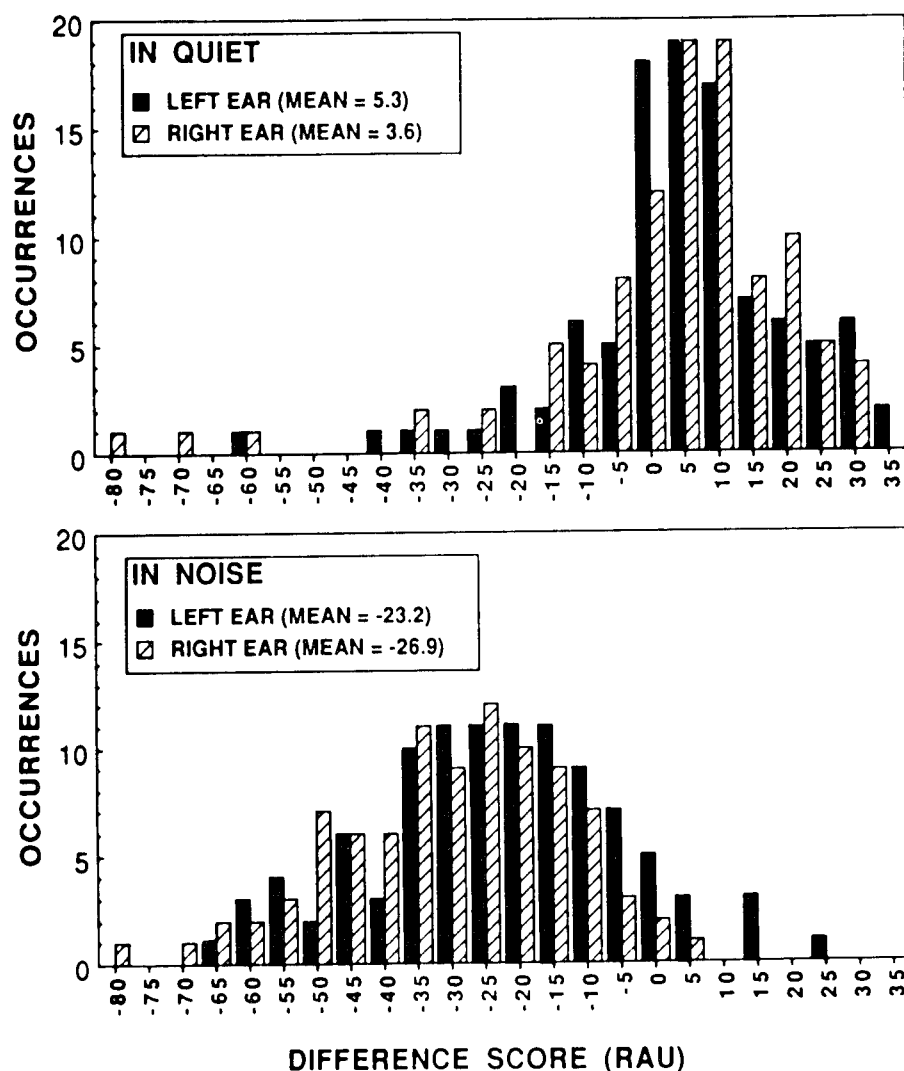


Figure 16-1 Distributions of difference scores based upon the application of the Articulation Index (AI) to speech perception measured in quiet and in the presence of a competing babble. The difference score is the difference between AI predictions and actual performance on NU-6 words in quiet and SPIN sentences in babble.¹

with perfectly normal auditory thresholds, Martin and Pickett showed that masked thresholds within the frequency region of the masker were identical, but, for signal frequencies above the noise, the range of individual differences was 35 dB. How could there be a range in off-frequency masked thresholds of 35 dB when quiet auditory thresholds and on-frequency masked thresholds were virtually

identical? Because of these and other data, and the obvious applications to individual differences in speech perception among normal hearing persons as well as the hearing impaired, we embarked on a study of upward spread of masking using a large group of human subjects with and without hearing loss. Because of a longstanding interest in the anatomical/physiological bases of this phenome-

non, we also replicated some of the conditions with experimental animals using the same low-pass filtered noise.

Human Psychophysical Data

Figure 16-2 shows an example of an unusual amount of upward spread of masking in a human observer.⁴ The masker was a low-pass filtered noise at 1.0 kHz at a level of 90 dB SPL. Quiet and masked thresholds were measured from 0.5 to 6.0 kHz. Note the systematic decline in masked thresholds for control subjects as the frequency of the test signal is increased from 1.5 to 6.0 kHz (filled circles). Open squares indicate the masked thresholds for this highly selected subject. Note that the on-frequency masked thresholds for this subject are identical to the control masked thresholds; however, from 1.5 to 3.5 kHz, there is substantial overmasking (10–25 dB) as indicated by the crosshatched area.

Figure 16-3 shows another example of an unusual amount of upward spread of masking.⁴ In this subject, it is particularly unusual because the masked thresholds off frequency

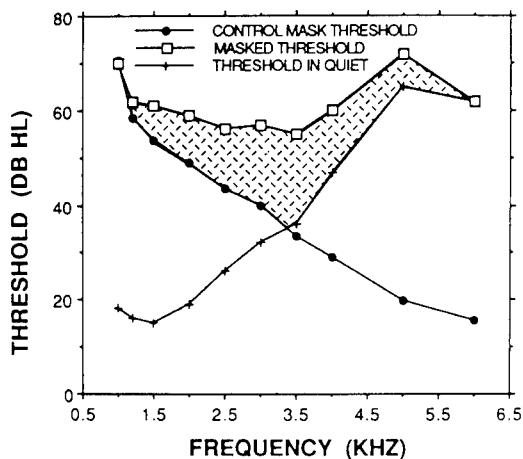


Figure 16-2 Example of overmasking (hatched area) in a human observer. Overmasking is defined as the region that exceeds the masked thresholds observed in young normal-hearing subjects or the quiet thresholds in the hearing-impaired subject.⁴

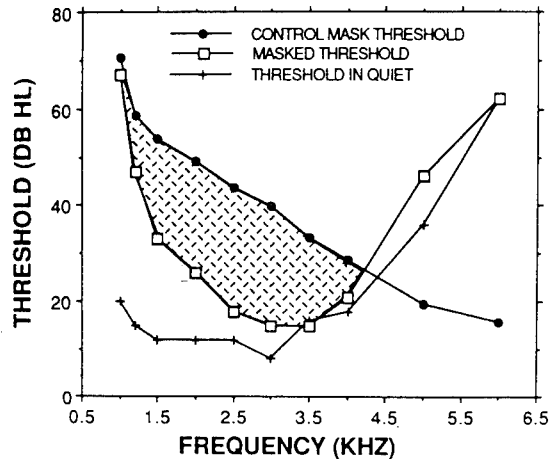


Figure 16-3 Example of undermasking (hatched area) in a human observer. Undermasking is defined as masked thresholds that are less than those observed in normal-hearing observers.⁴

are 22 dB better than observed in control subjects. The crosshatched area indicates the magnitude of the undermasking effect. In other words, Figure 16-2 shows an example of too much masking whereas Figure 16-3 shows an example of too little masking. It is important to note that on-frequency masking is essentially identical to that of control subjects and to the experimental subject of Figure 16-2. It is also important to note that at the 2.0 kHz area, for example, the range of masked thresholds is about 40 dB. In other words, we observe on-frequency masked thresholds that are virtually identical and predictable, and off-frequency masked thresholds that differ by 40 dB and are unpredictable.

The undermasking and overmasking phenomena shown in Figures 16-2 and 16-3 are clearly correlated with speech perception in noise. Figure 16-4 shows the masking of speech plotted as a function of the upward spread of masking as defined as the difference in masked thresholds between 1.0 and 2.0 kHz.⁵ Persons with shallow masking slopes are "overmasking subjects" whereas those with steep slopes are "undermasking subjects." Correlations between masked speech thresh-

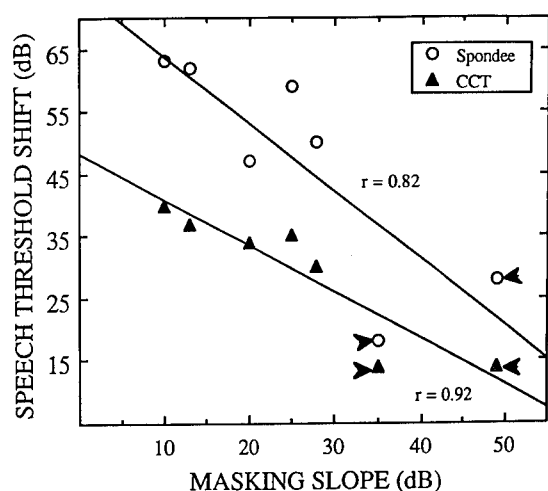


Figure 16-4 The speech threshold shift (masking) as a function of the slope of upward spread of masking defined as the difference in masked pure-tone thresholds at 1.0 and 2.0 kHz. (►◄) Subjects with undermasking; (○) spondaic words; and (▲) results of the California Consonant Test.⁵

olds in a low-pass masker and the slope of the masking function are highly correlated, namely $r = 0.82$ for spondaic words and $r = 0.92$ for the California Consonant Test.

Mean absolute and masked psychophysical thresholds obtained from four groups of human subjects⁴ are shown in Figure 16-5. Results for young and aged groups of normal-hearing subjects are shown in the top panel. Note that absolute thresholds for the aged subjects are nearly equal to those of the young subjects (left panel). In addition, thresholds measured in a low-pass masker (right panel) are nearly identical for the two age groups, for signals within the masker passband (at 1.0 kHz) and outside the passband. Similarly, absolute thresholds for young and aged groups of hearing-impaired subjects (in the bottom left panel) show similar magnitudes of hearing loss, as well as similar thresholds measured in a low-pass masker (bottom right panel). In addition, it is clear that masked thresholds for hearing-impaired subjects are higher than for normal-hearing subjects. As such, differences in masked thresholds between young and aged hearing-impaired subjects are difficult to interpret due to the influence of absolute

thresholds on masked thresholds. Using a masked-threshold prediction scheme it was determined that the magnitude of overmasking both within and outside the masker passband (i.e., at 1.0, 2.0, and 3.0 kHz) was not significantly different for young and aged hearing-impaired subjects. Thus, magnitude of hearing loss, rather than age, appears to be the major contributing factor in human psychophysical masked thresholds.

An effort using human subjects to identify possible physiological correlates of upward spread of masking, including both overmasking and undermasking, were largely unsuccessful. Thresholds of the auditory brain stem response (ABR) suggested that the effect occurred at sites peripheral to the generation of the wave V potential. Measures of acoustic distortion products were inconclusive. In light of these data and other considerations, we attempted to develop an animal model of masking using the Mongolian gerbil.

Animal Studies of Masking

Figure 16-6 shows unmasked and masked thresholds for young gerbils with normal hearing and aged gerbils (36 months of age).⁶ Both groups were born and reared in a quiet vivarium. Auditory thresholds were estimated using ABR recorded first when the animal was 4–6 months of age, and then at periodic intervals. At age 36 months, final ABR and other physiological measures were made and the animal prepared for anatomical studies. The ABR procedure is nearly fully automated. The gerbil is sedated with ketamine/xylazine, needle electrodes are attached at the vertex and mastoids, and tone bursts from 1.0 to 16.0 kHz are presented at levels from 10 to 80 dB sound pressure level (SPL). The tone burst is 1.8 milliseconds in duration. Spectra of the auditory signals as well as more details of the ABR procedure are given elsewhere.⁷

In Figure 16-6 (left panel) the unmasked thresholds of the young and aged animals differ by as little as 12 dB at 2.0 kHz to as much as 27 dB at 8.0 kHz. This age-related loss of hearing is consistent with our earlier data.⁷ The

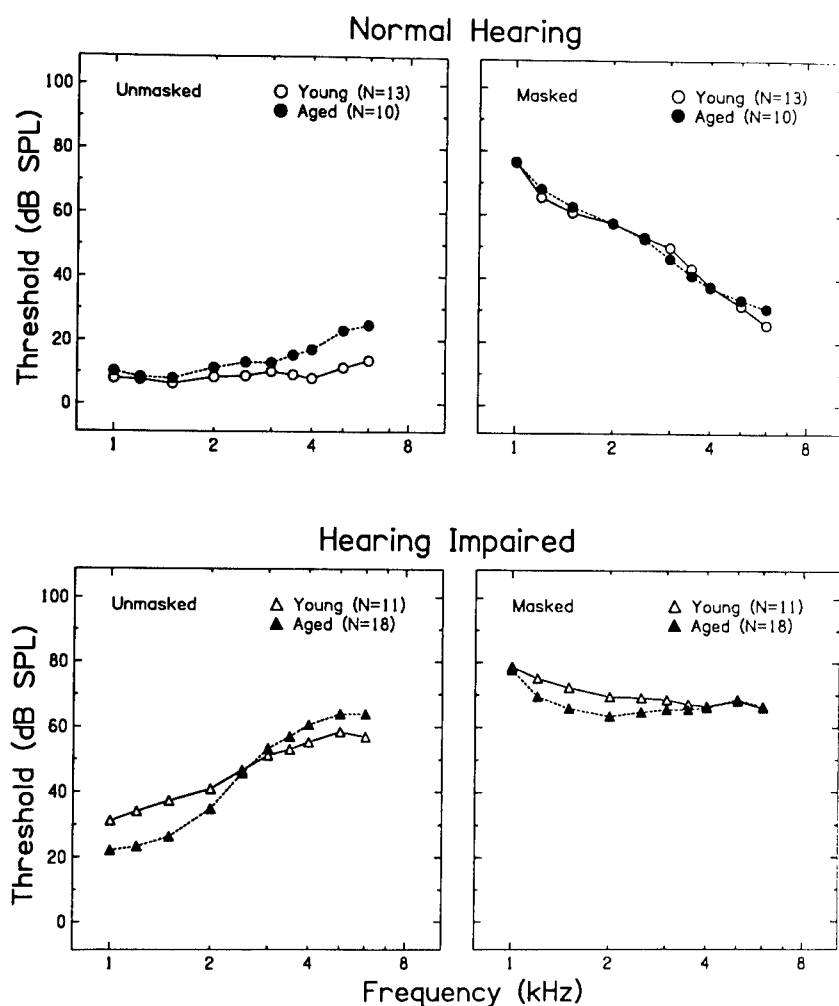


Figure 16-5 Absolute and masked thresholds for young and aged subjects with normal hearing and hearing impairment. The data show very little difference, if any, between young and aged subjects. The major variable is hearing loss as indicated by the absolute threshold.⁴

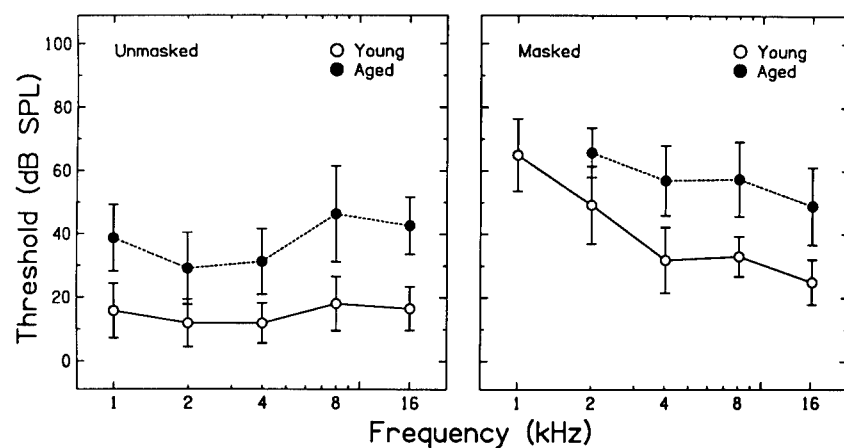


Figure 16-6 Unmasked and masked ABR thresholds in young and 36-month-old gerbils. Vertical bars indicate ± 1 standard deviation.⁶

vertical bars indicate ± 1 standard deviation. Note that the variance of the aging animals is substantially greater than that observed for the young animals. The right panel of Figure 16-6 shows masked thresholds where the masker was a noise low-pass filtered at 1.0 kHz with a level of 80 dB SPL, and a high-frequency roll-off of 96 dB/octave. The masked thresholds for the young animals show a predictable amount of masking at 1.0 kHz (on frequency) and a systematic reduction in masked thresholds as the frequency of the test tone is increased to 4.0 kHz. For the aging animals, a masked threshold at 1.0 kHz could not be obtained for many of the animals at the maximum output of the acoustic system, that is, 75–80 dB SPL. At 2.0 and 4.0 kHz, masked thresholds were substantially higher than those of young animals. Likewise, the slope of the masking function between 2.0 and 4.0 kHz is much steeper in the younger animals than in the older animals. The steeper slope as well as the much larger masked thresholds in the aging animals is consistent with the definition of overmasking. It is important to note, however, that we were unable to obtain masked thresholds at 1.0 kHz (on frequency) in many of the aging animals. This result is a complicating and potentially important difference between the masked thresholds we have observed in our human subjects (Figure 16-5) and the masked thresholds observed in our aging animals. Also, we do not believe we have observed the equivalent of undermasking in our aging animals.

Do the higher masked thresholds (Figure 16-6) in the aging animals reflect overmasking, that is, an unusual upward spread of masking? Or do the differences in masked thresholds between the young and aging animals simply reflect the differences in auditory sensitivity between the two groups (as shown in the left panel of Figure 16-6)? To test this notion, that the differences in overmasking simply reflects differences in auditory sensitivity, we selected two aging subjects with normal or nearly normal auditory sensitivity and compared their masked thresholds to worst-case predictions of masked thresholds for young animals with normal hearing. Fig-

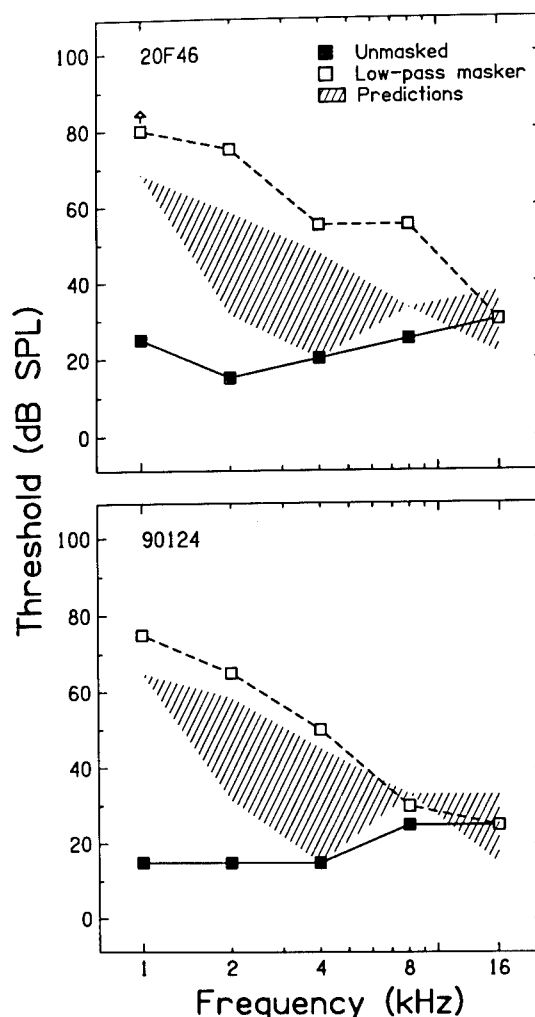


Figure 16-7 Unmasked and masked ABR thresholds for two (individual) 36-month-old gerbils and range of predictions of masked thresholds (hatched area) using the modified power law of Humes and Jesteadt.⁸

ure 16-7 shows the results. The crosshatched area indicates the range of masked thresholds for normal animals as predicted by the modified power law.⁸ Clearly, the masked thresholds for the aging animals are greater than those observed for the worst-case normal young animals. Of course, if we used the 99% confidence interval for normal young animals, the differences between masked thresholds of our two aging animals with excellent hearing and the control subjects would be even

greater. It is thus quite clear that masking of the ABR by a low-pass masker is significantly different in young and aged animals who are raised in a quiet environment, and that this difference is not explainable in terms of hearing loss as indicated by a loss of auditory sensitivity.

Using many of the aging animals who had participated in the studies of masking of the ABR by a low-pass noise and who had demonstrated excessive upward spread of masking, one of our colleagues (Schmiedt, as reported in Boettcher et al.⁹) studied masking of the compound action potential (CAP) of the auditory nerve. Using a round window electrode and a masker and signals identical to those used in the ABR studies, Schmiedt found masked CAP thresholds that were virtually identical to masked ABR thresholds. Thus, the overmasking observed in aging gerbils is observed in the auditory nerve as well as in the auditory brain stem, and thus has its origins in the auditory periphery.

Frequency Selectivity in Quiet-Aged Animals

There are several acceptable criteria of frequency selectivity or the frequency resolving power of the ear. In the psychophysical domain, the most common measures are differential sensitivity for frequency, critical ratios or signal-to-noise ratio at masked threshold, critical bandwidth as assessed in masking experiments, psychophysical tuning curves, and a number of variations on these basic measures. In keeping with the traditional measures of frequency selectivity, unusual upward spread of masking as produced by a low-pass filtered noise would be an example of a loss of or a degradation in frequency selectivity. In other words, abnormal masking patterns are an indication of an abnormal ear with respect to frequency selectivity.

In the physiological domain, the most basic and common measure of frequency selectivity is the tuning curve recorded from a single fiber of the auditory nerve. From the tuning curve (Figure 16-8,¹⁰ thick line) two factors describe the sharpness of the tuning. One is the high-

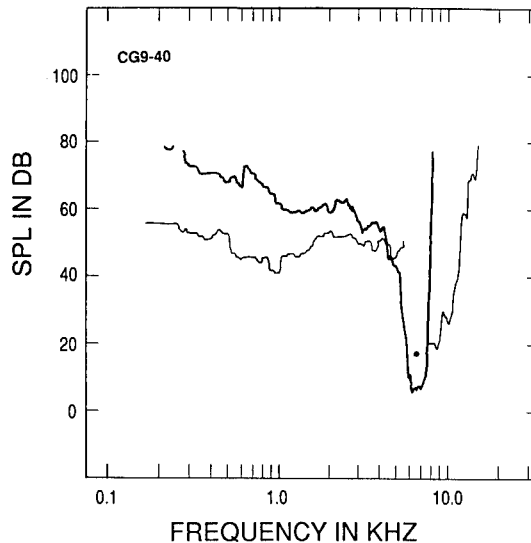


Figure 16-8 Tuning curve and areas of two-tone rate suppression from a single nerve fiber of the auditory nerve of a young gerbil.¹⁰

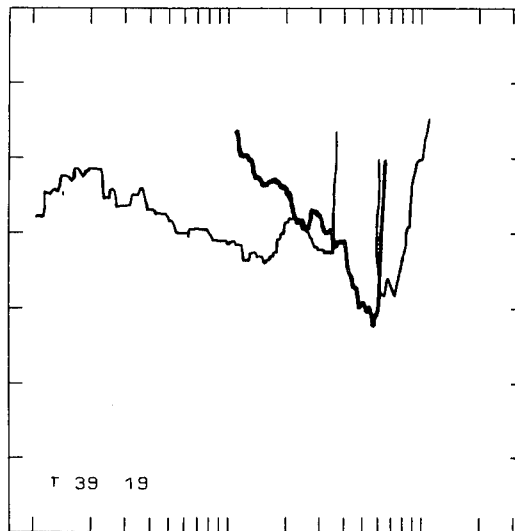


Figure 16-9 Tuning curve and areas of two-tone rate suppression from a single nerve fiber of the auditory nerve of a 36-month-old gerbil. Note that the sensitive tip of the tuning curve is reduced, but the slopes of the tuning curve are sharp, and most importantly, two-tone rate suppression is present both on the low- and high-frequency side of the tuning curve.¹⁰

frequency side of the tuning curve that is very steep, about 200 dB/octave or greater. The second is the bandwidth of the tuning curve that is arbitrarily taken as the tuning-curve bandwidth at a point 10 dB up from the tip of the tuning curve ($Q_{10\text{dB}}$).

A second factor that contributes to the tuning of a single fiber involves a nonlinear phenomenon, two-tone rate suppression (Figure 16-8, thin line). In Figure 16-8, two-tone suppression is shown both above and below the characteristic frequency of the unit. We will not go into the basis of two-tone rate suppression or review the voluminous literature available. Rather, we wish to note that in ears that have been affected by exposure to noise or to ototoxic drugs, one of the first (if not the first) indicator of temporary or permanent injury is

the reduction or complete loss of two-tone suppression. Following the loss of two-tone suppression is a decrease in the slope of the high side of the tuning curve with an attendant increase in the bandwidth of the tuning curve. These changes in high-frequency slopes and bandwidth precede losses of auditory sensitivity that are indicated by an elevation of the tip of the tuning curve.¹¹

Given these observations on tuning curves and on two-tone suppression, the loss of auditory sensitivity in older animals, and the severe loss of frequency selectivity as indicated by the excessive amount of masking of the CAP and ABR in older animals, we fully expected to see tuning curves from single units of the auditory nerve of older animals to have elevated tips, shallow slopes on the high-

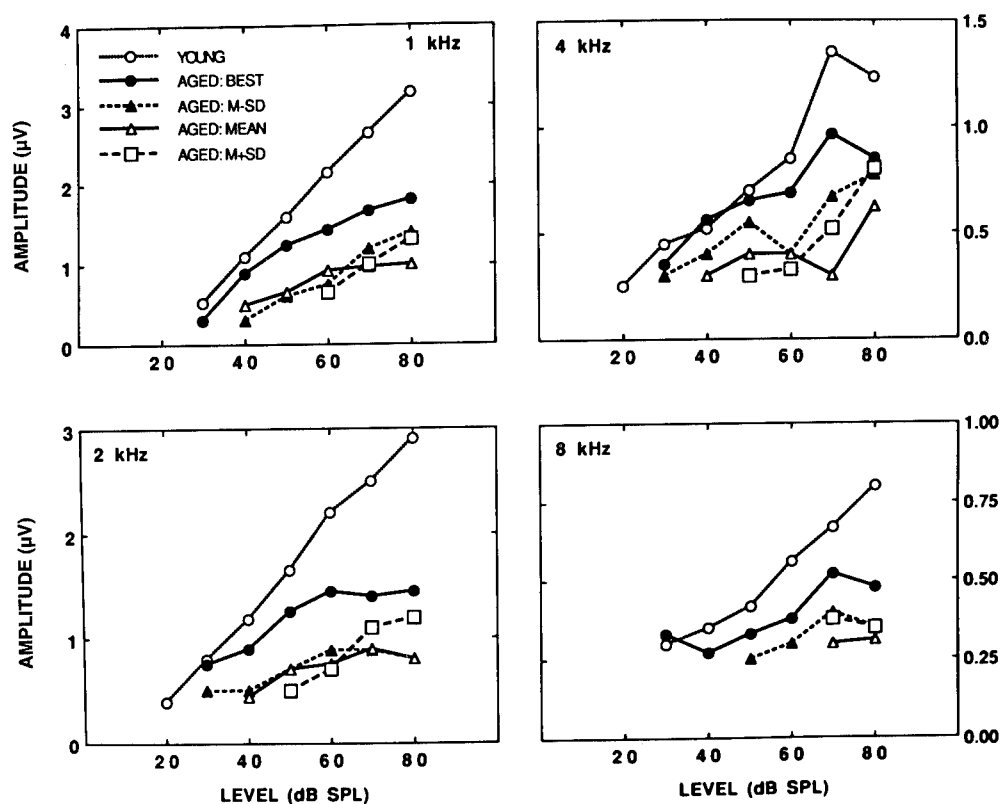


Figure 16-10 Amplitude-intensity function of wave IV of the gerbil ABR for tone bursts of 1.0, 2.0, 4.0, and 8.0 kHz. On each panel animals are grouped by hearing loss (mean, -1 SD, $+1$ SD, best) and compared to young animals.¹² Note that the slopes of the input/output functions are shallower than those of the young animals, even in some instances where there is little difference between the young and aged animals in terms of auditory sensitivity.

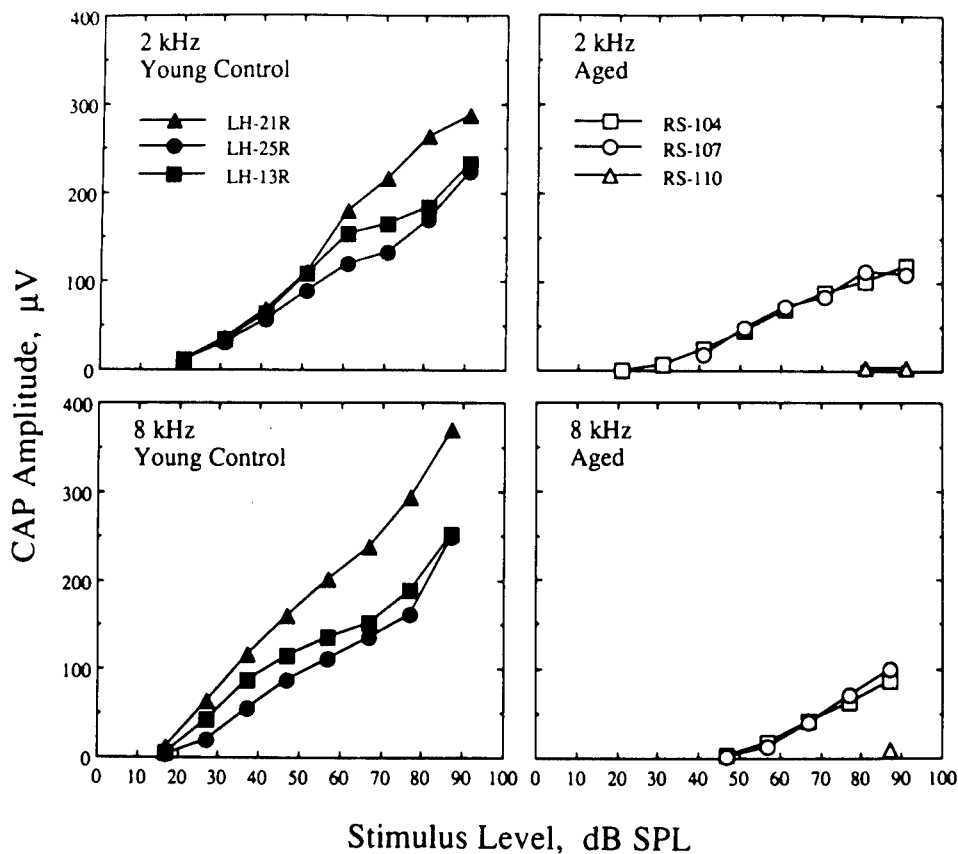


Figure 16-11 Amplitude-intensity functions of the compound action potential of the auditory nerve recorded near the round window in young and 36-month-old gerbils.¹³

frequency side, increased bandwidth, and an absence of two-tone rate suppression.

Figure 16-9 shows a tuning curve and two-tone rate suppression recorded from a single fiber of the auditory nerve of a 36-month-old gerbil.¹⁰ The tip is elevated about 30 dB, which corresponds to the hearing loss measured in many 36-month-old gerbils. This result is clearly expected. However, although the tip of the tuning curve is elevated, both the low- and high-frequency sides of the tuning curve appear normal or nearly so, bandwidth is normal or nearly so, and most importantly, two-tone rate suppression is present and robust on both the high- and low-frequency sides. These observations were not expected and are clearly inconsistent with the CAP and ABR masking data. In other words, we have a pronounced decrease in frequency selectivity as indicated

by the CAP and ABR in the same animal, but we have single nerve fibers that are normal with respect to the usual and customary measures of frequency selectivity.

It is our working hypothesis that a loss of auditory sensitivity accompanied by a loss of tuning including suppression at the level of the auditory nerve is an indication of injury to the hair cell system. This type of injury is commonly produced by most ototoxic agents including noise. In the aging animal on the other hand, it is hypothesized that the hair cell system is largely intact and that the major pathology is in the lateral wall system including the stria vascularis. Thus, as Schmiedt et al.¹⁰ have suggested, noise and drugs play havoc with cochlear micromechanics, whereas aging per se may have most of its effects on the highly active ion pumps of the lateral wall.

A potentially confounding factor in the overmasking data observed at the level of the CAP and ABR is the unusual input/output functions of the CAP and ABR. Figure 16-10 shows input/output function of wave IV of the gerbil ABR.¹² Note in Figure 16-10 that even for animals with minimal if any hearing loss, the slope of the input/output function is much shallower than that of young controls. In addition, the maximum amplitude of the ABR is always considerably smaller than that of young controls. Figure 16-11 shows CAP input/output functions for young and aged animals that are consistent with ABR data, namely, input/output

functions for older animals with essentially normal hearing have shallower slopes and the potentials have much lower amplitudes.¹³

Input/output functions recorded from single fibers of the auditory nerve of aging gerbils (Figure 16-12) are inconsistent with input/output functions of the CAP (Figure 16-11) and ABR (Figure 16-10). In aged animals, the shape of the input/output function, the normal saturation rates, and the fact that spontaneous rates do not fatigue,¹⁴ all suggest that the hair cell/nerve fiber synapse is normal. This is, of course, in contrast to the effects of ototoxic drugs and noise.

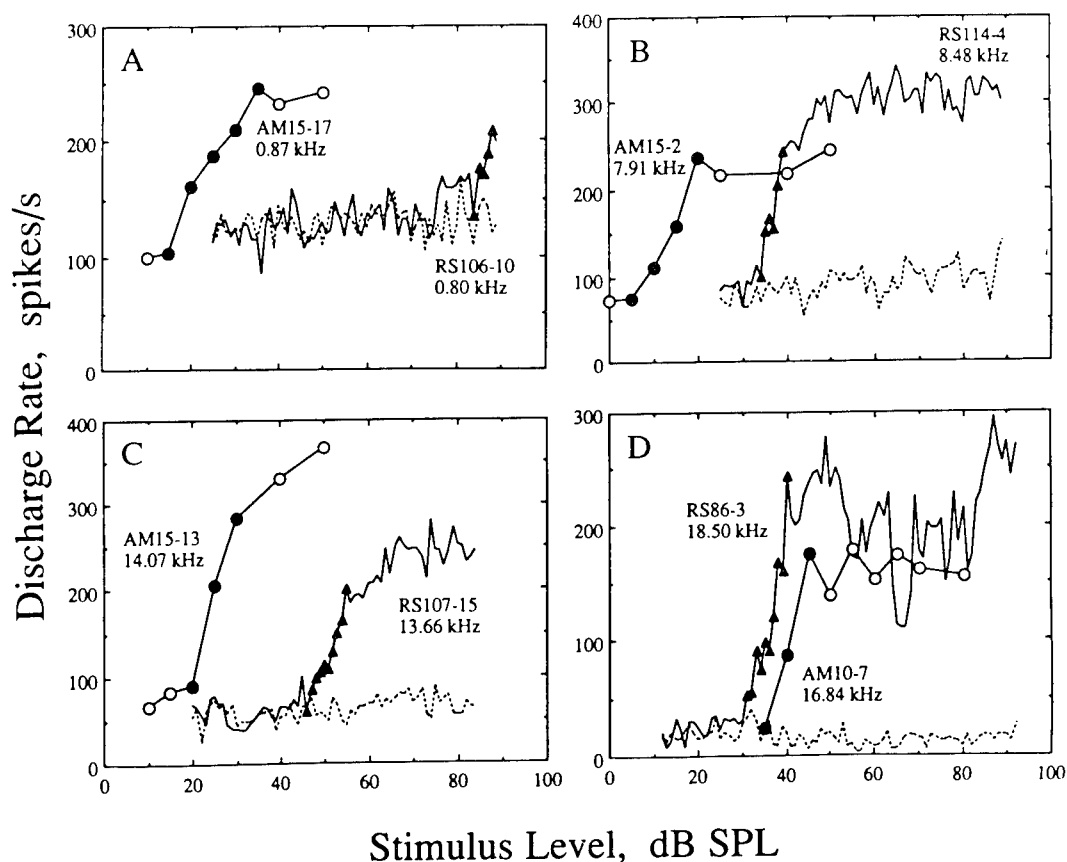


Figure 16-12 Steady-state, single-unit rate/level functions for (○, ●) young controls and (—▲—) 36-month-old gerbils. (· · ·) Nondriven (spontaneous) activity.¹⁴ Rate/level functions of the aged animals are very similar to young controls in terms of slope and saturation, but have thresholds that are shifted to higher stimulus levels. This result is consistent with single-unit tuning curve data, but is inconsistent with slopes of amplitude/intensity functions of both the CAP and ABR, and masked CAP and ABR thresholds.

It would thus appear that single-fiber rate/level functions do not contribute to the decreased slopes of the CAP or ABR input/output functions. Likewise, it would seem that single fiber tuning curves and the non-linear mechanisms that produce two-tone rate suppression contribute little to the decrease in frequency selectivity observed in both the CAP and ABR of aging animals. Two possibilities remain. One possibility is a lack of synchrony in the responses of aged auditory-nerve fibers. Whereas this is an extremely likely possibility, we have no data to support it. Loss of myelin sheaths around nerve fibers could contribute significantly to a lack of synchrony as well as to a loss of tuning. A second factor is a loss of auditory-nerve fibers in aging animals.^{15,16} The number of fibers with low and medium spontaneous rates (below 18 spikes/s) are proportionally fewer in aging animals.¹⁷ Because these fibers typically have higher thresholds than fibers with higher spontaneous rates, the loss of these lower spontaneous rate fibers might not allow a complete growth of the CAP input/output function. In turn, the reduced output from the cochlea is reflected in ABR input/output functions. Thus, it is our current working hypothesis that the decreased slopes of the CAP and ABR input/output functions are attributable to a loss of synchronized activity in the responses of auditory-nerve fibers, and to a decrease in the number of spiral ganglion cells in aging animals. Although these two factors may be able to explain abnormal input/output functions at the level of the auditory nerve and brain stem, additional data are needed on the pathologic anatomy and physiology associated with unusual amounts of psychophysical masking, that is both overmasking and undermasking, observed in human subjects, and abnormal masking observed in the CAP and ABR of aged gerbils.

Acknowledgments

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Chapter 17

Interactions Between Age-Related and Noise-Induced Hearing Loss

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and Flint A. Boettcher

The most common causes of sensorineural hearing loss are exposure to noise, aging, and the interactive effects of exposure to noise and aging. One of the longstanding issues in the assessment of noise-induced hearing loss is the interaction between hearing loss produced over a working lifetime in occupational noises and the hearing loss associated with increased chronological age and other factors. The accurate assessment of hearing loss in older persons, primarily for medicolegal reasons, requires the allocation of hearing loss into the components associated with increased chronological age and other factors. Sometimes there is also the need to assign the hearing loss to a single source of occupational noise or to repeated exposures to a wide variety of occupational noises. Although perhaps seemingly straightforward, the task faced by the professional is potentially quite complicated and perhaps not even possible in the absence of accurate, quantitative, noise exposure data and longitudinal audiometric data.

One longstanding approach is based on population studies of individuals with known occupational noise exposures (experimental groups) and of individuals with no history of occupational noise exposure (control groups). Experimental and control groups are assumed to have age-related hearing losses that are equal in all respects, genetically determined, and "contaminated" by the insidious effects of sociocusis (nonoccupational exposures to noise) and nosocausis (diseases of the ear that are largely undiagnosed and unassessed). It is

assumed that the hearing loss that is age-related adds to the hearing loss produced by the occupational noise exposures. Thus, the traditional approach assumes age-related hearing loss adds (in decibels) to the noise-induced hearing loss. The additivity (more accurately, multiplicative) assumption is supported by epidemiologic data and is embraced by the recent international standard (ISO 1999).¹ In ISO 1999, a compression factor is introduced and becomes meaningful for hearing losses over about 40 dB. Much debate and uncertainty remains about the accuracy of both experimental and control epidemiological data and about the potential limitations in applying epidemiological data that has been averaged across thousands of persons to individual persons about whom very little is known.

Laboratory Studies With Animals

In light of the complexities and potential limitations of epidemiologic studies, there is a need for an alternative approach to issues in noise-induced hearing loss. Our approach is the use of an animal model (Mongolian gerbil) in studies of presbycusis, noise-induced hearing loss, and the interaction of the two. The gerbil model allows the identification and control of the most pertinent variables as well as systematic studies of many of the empirical issues. In one of our first studies^{2,3} we compared hearing loss produced in gerbils who spent most of their lives in an 85 dBA sound field with hearing losses produced in gerbils

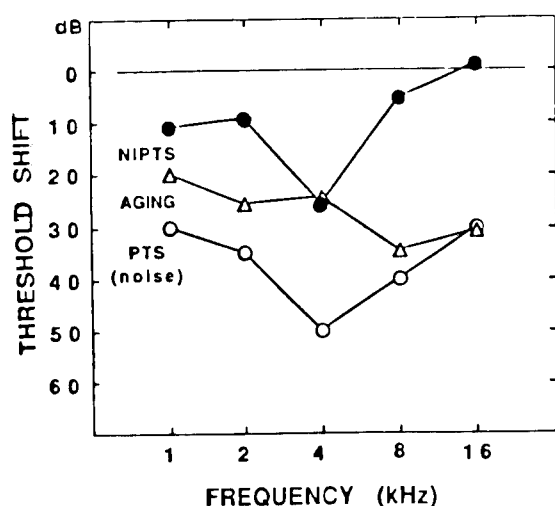


Figure 17-1 Threshold shifts in dB in 36-month-old gerbils. PTS is the median permanent threshold shift in seven gerbils who were exposed for 24 h/d over 2 years to a wide band noise at 85 dBA. Aging refers to the mean threshold shift observed in gerbils ($n = 32$) who were born and reared in quiet (41 dBA) animal quarters. NIPTS is a difference curve where the aging median data are subtracted (in dB) from the PTS median data. Auditory thresholds were estimated from ABR measurements. (After Mills et al.^{2,3})

who were born and reared in a quiet vivarium (about 41 dBA). Figure 17-1 shows the permanent threshold shifts (PTS) in the noise-exposed animals after about 720 days of exposure. Also shown in Figure 17-1 are threshold shifts of aging control animals. Thresholds and threshold shifts are estimated from electrical potentials arising from the auditory brain stem. These auditory brain stem responses (ABR) are obtained with the animal anesthetized. At a test frequency of 16 kHz, ABR thresholds are equal for noise and control subjects. At 8 kHz the hearing losses of the noise group are only 5 dB greater than those of the controls. In the frequency region encompassed by the noise, 500–4000 Hz, hearing losses of the noise-exposed animals clearly exceed those of the controls. In other words, hearing losses at 8 and 16 kHz in the noise-exposed animals were probably attributable to the effects of aging rather than to the effects of exposure to noise. At test frequencies of 1, 2,

and 4 kHz it remains unclear in the noise-exposed animals how much of their hearing loss should be attributed to the noise exposure and how much should be attributed to the effects of aging. In Figure 17-1 we have simply subtracted the median hearing loss of the aging noise-exposed animals from the median hearing loss of the aging control animal, to obtain the difference curve of noise-induced PTS (NIPTS). This “subtraction technique” corresponds to the method used to correct audiometric surveys of persons exposed to noise, that is, the hearing loss of the noise-exposed group is “corrected” by subtracting the mean or median hearing loss of a control group. The shape of this difference curve in Figure 17-1 corresponds closely to the spectrum of the noise. This correspondence supports the notion that the difference curve is truly attributable to the noise exposure and supports the additivity methods used in the estimation of NIPTS.

For the group data of Figure 17-1 the residual (NIPTS) hearing loss appears logical, consistent with a priori expectations, and supports the concept of additivity of age-related hearing loss and noise-induced hearing loss; however, when hearing losses of individual animals are examined (Figure 17-2), the situation is clearly not as straightforward as one would assume from Figure 17-1. Hearing losses were averaged across test frequencies 1, 2, 4, and 8 kHz, and the distribution of threshold shift for the noise-exposed animals and quiet-reared animals is depicted in Figure 17-2. It is noteworthy that threshold shifts of two control animals exceed the threshold shifts of all of the noise-exposed animals. Indeed, the variability among control animals (born and reared in a quiet vivarium) is greater than the variability of the animals who spent most of their lives in an 85 dBA noise field. Clearly, variance among control animals complicates the assessment of NIPTS in individuals, even under experimental conditions where it is reasonable to assume that the significant variables were under the control of the experimenter.

Another indication of variability in control animals is shown in Figure 17-3. These animals have been grouped in Figure 17-3 by

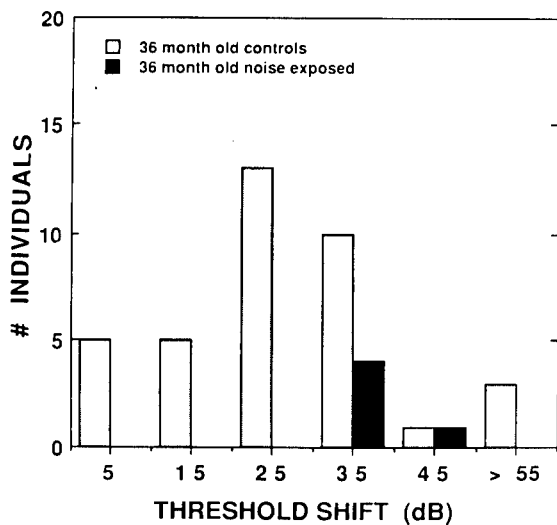


Figure 17-2 Distribution of the threshold shift data for the 36-month controls ($n = 37$ ears) and 36-month noise-exposed animals ($n = 7$) from Figure 17-1. Note that the animals with the poorest hearing were not in the noise-exposed group, but were from the group born and reared in a quiet vivarium. (After Mills et al.²)

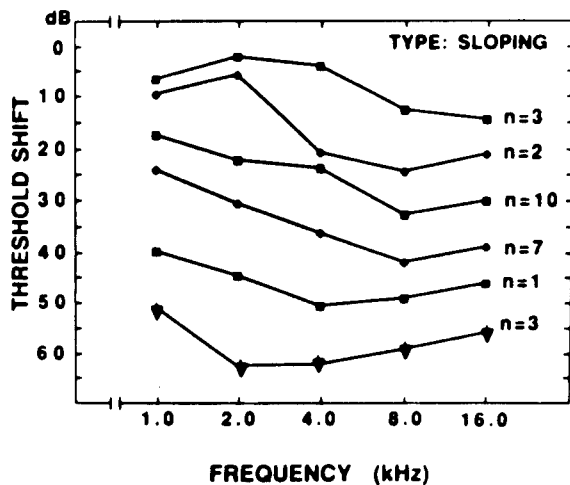


Figure 17-3 Some of the aging data from Figure 17-1 have been grouped by the similarity of the audiometric configuration. Note that hearing losses, averaged from 1 to 16 kHz, range from nearly 0 to 5 dB for three animals to >65 dB for three other animals. (After Mills et al.³)

audiometric similarity. Note that for the 26 ears depicted in Figure 17-3, threshold shifts range from nearly 0 dB to greater than 70 dB. Variance of this magnitude is remarkable given that the chronological age, environment, acoustic history, and diet is virtually identical for all animals. Moreover, in addition to control of the acoustic environment, none of the animals had any history of drug administration. Animals with conductive hearing loss were eliminated.

Two factors emerge from the aging-noise gerbil data of Figures 17-1 to 17-3. One is the presence of a strong genetic factor in age-related hearing loss in the gerbil. In our colony the inbreeding coefficient is 0.12. If through selective breeding we raised this coefficient to 0.9, we would have an inbred colony and the variance could be hypothetically reduced to nearly 0. Likewise, we could reduce the coefficient of inbreeding to nearly 0 and thus increase the variance of quiet-reared animals even more. Efforts to systematically control the variance in our aging animals and to breed presbycusis resistant and presbycusis susceptible animals are very expensive and time consuming although potentially rewarding.

A second factor that emerges from the aging data of Figures 17-1 to 17-3 is that the quantitative differentiation of noise-induced hearing loss from age-related hearing loss is a formidable (if not nearly impossible) task. This is true even for Mongolian gerbils who are raised in controlled environments, and for whom accurate, longitudinal noise exposure and audiometric data are available. Of course, this information is often not available for humans.

Epidemiologic Data

The effects of aging clearly complicate the assessment of noise effects in the Mongolian gerbil model as well as complicating the situation with humans. Indeed, for noise exposures with levels of 100 dBA and less, age-related PTS (ARPTS) is the principal component of hearing loss measured in persons who have been exposed for 8 h/d for 40 years.⁴ It is also difficult to separate effects that are related to exposure duration rather than to age-related

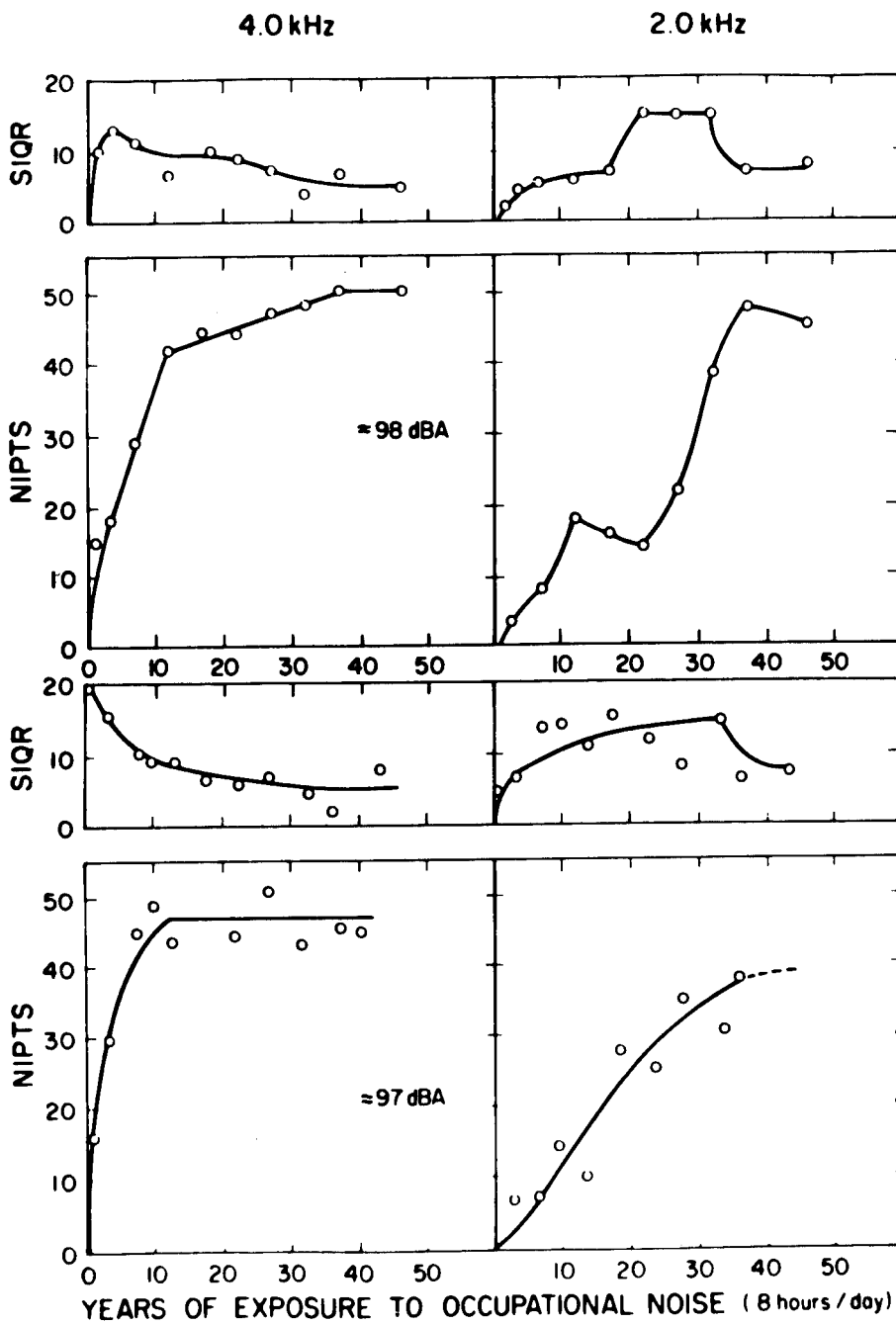


Figure 17-4 Human NIPTS data from Taylor et al.⁶ (top half), and Nixon and Glorig⁵ (bottom half). NIPTS at 4 kHz is shown on the left-hand side of the figure and 2 kHz is shown on the right-hand side. Variability is given by the semiinterquartile range, SIQR. (After Mills⁷)

hearing loss. For example, individual differences in NIPTS, that is, PTS corrected for the effects of aging, appear to increase and then decrease as an exposure is continued over years. Part of this temporal effect may reflect the aging process, but other portions of this temporal effect occur after a few years of exposure where aging effects are minimal to nonexistent.

Changes in the variability of NIPTS as a function of exposure duration are shown for human workers in Figures 17-4 and 17-5. Nixon and Glorig⁵ reported median and quartile values of NIPTS at 4.0 and 2.0 kHz for exposures with A-weighted sound levels of approximately 83, 92, and 97 dB, and durations of up to 30 years (8 h/d; 5 d/w). Taylor et al.⁶ reported similar data but only for one level of noise, about 100 dBA. Figure 17-4 depicts

the Taylor et al. data at 4.0 and 2.0 kHz, and the Nixon and Glorig data at 4.0 and 2.0 kHz for their 97 dBA condition. The abscissas of Figure 17-4 are the durations of exposure in years. The ordinates show median NIPTS and the variability of NIPTS as indicated by the semiinterquartile range (SIQR).⁷ At 4.0 kHz note that the median NIPTS increases rapidly during the first 10 years of exposure and then is asymptotic as the Nixon and Glorig⁵ data suggest (lower half of Figure 17-4) or increases slightly between 20 and 50 years of exposure as the Taylor et al.⁶ data suggest (top half of Figure 17-4). Note also that at 4.0 kHz the variability of NIPTS as indicated by the SIQR reaches a maximum value of 12–20 dB after a few months (Nixon and Glorig data⁵) or a few years (Taylor et al.⁶) and then decreases systematically to an asymptotic value of about

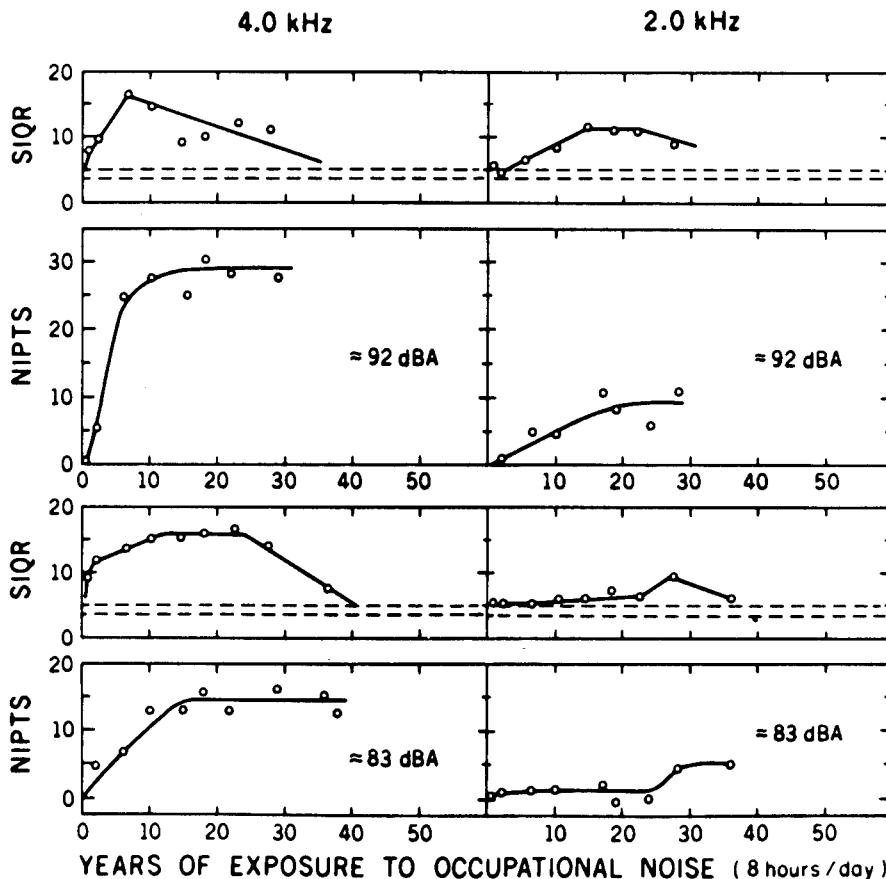


Figure 17-5 Data as in Figure 17-4 except that the data are only from Nixon and Glorig.⁵ (After Mills⁷)

5 dB. It is quite possible that this asymptotic value of 5 dB is equal to the preexposure SIQR.

At 2.0 kHz the median NIPTS increases for about 30 years of exposure and then appears to reach an asymptote. Variability of NIPTS as derived from the Taylor et al.⁶ data reaches a maximum after 20 years of exposure, remains constant between about 20 and 30 years of exposure, and then decreases and reaches an asymptotic value of about 5–7 dB after about 40 years of exposure.

Variability at 2.0 kHz as indicated by the Nixon and Glorig⁵ data is less systematic than the data of Taylor et al.⁶ In this case, variability increases at least for the first 10–20 years of exposure and possibly for the first 35 years of exposure. Thereafter, as in the Taylor et al. data,⁶ variability decreases to an asymptotic value of about 5–7 dB.

The major difference between variability at 4.0 and 2.0 kHz appears to be in the temporal domain. That is, the maximum SIQR at 2.0 kHz (16–17 dB) is nearly equal to the maximum at 4.0 kHz; however, the maximum at 2.0 kHz is reached after nearly 20 years of exposure whereas the maximum at 4.0 kHz is reached after less than 5 years of exposure. This difference in the time domain between 4.0 and 2.0 kHz is most apparent for the 97 and 98 dBA exposures where the median NIPTS is greater than 15 dB.

Figure 17-5 shows data from Nixon and Glorig⁵ as in Figure 17-4 except that the noise exposure levels are 92 dBA (upper panel) and 83 dBA (lower panel). The suggestions about NIPTS and its variance as shown in Figure 17-4 for exposures of about 100 dBA are also true for Figure 17-5 for 92 and 83 dBA exposures with one noteworthy exception. At a test frequency of 4 kHz and an exposure level of 83 dBA, the variability is larger than in any other exposure condition, and reaches a maximum value after 25–30 years of exposure. Indeed, the SIQR at 4 kHz after 30 years of exposure at 83 dBA is about 16 dB, whereas the SIQR at 4 kHz after 30 years of exposure at 100 dBA is only 8 dB. One interpretation of this result is that the 100 dBA exposure is so dominant that individual differences due to

noise and other factors (including aging) are diminished. In other words, the noise exposure is so severe it makes everyone “nearly equal.” For the 83 dBA exposure, on the other hand, the noise exposure produces only minor changes in hearing. Other factors including age-related threshold shift are dominant.

Perhaps the most significant feature is that the variability of NIPTS always reaches a maximum several years before the median NIPTS and at a time when the rate of change of the median is greatest (or nearly so). Similarly, variability of NIPTS is smallest when the median NIPTS is largest and the rate of change of the median is minimal. In other words, the relation between the variability of NIPTS and the median NIPTS is nearly identical to that observed in many dynamic systems where the variability of response varies directly with the rate of change of response.

Laboratory Studies With Humans

Time-dependent variability in noise-induced threshold shifts is not restricted to field studies. Figure 17-6 shows temporary threshold shifts (TTS) data for four individuals exposed for 24 hours to a band of noise. Note that after 1 hour of exposure there is a 14 dB difference between the subjects; after 24 hours the difference is only 7 dB. In other words, significant differences between subjects early in an exposure can pale to insignificance as the exposure is continued. Clearly, these temporal variations are unrelated to the aging process. In the industrial situation, individual differences diminish after many years of exposure and at a time when aging effects are clearly becoming operational, and at test frequencies of 3, 4, and 6 kHz after about only 10 years of exposure and before aging effects are easily measured. In this latter field situation and in the laboratory studies of TTS, the anatomical–physiological bases of these temporal variations is not at all clear. In any event, individual differences are a significant and complicating variable in noise-aging studies in humans just as they are in noise-aging studies in experimental animals.

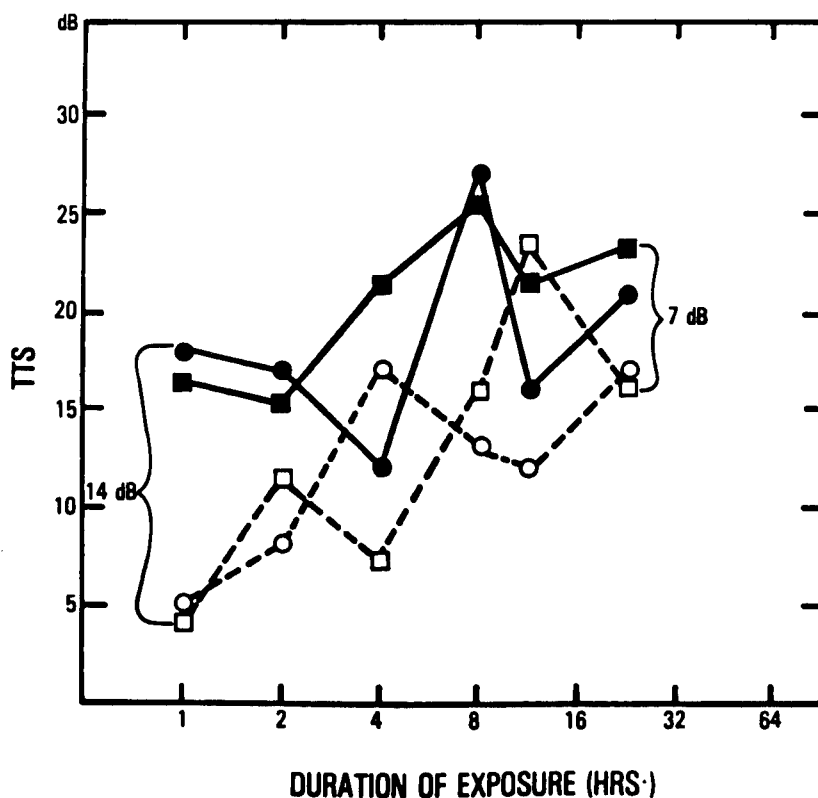


Figure 17-6 Temporary threshold shift at 4 kHz in four individuals for exposure durations ranging from 1 to 24 hours. Seemingly large differences between individuals early in an exposure are not sustained as the exposure is continued (J.H. Mills, unpublished data, 1995).

Allocation of Hearing Loss

For many years there has been an interest in the assessment of noise-induced hearing loss in individuals for several reasons, including claims for workers' compensation. Accordingly, there is an interest in separating or differentiating the hearing loss caused by occupational noise(s) from the hearing loss associated with sociocausis and nosocausis. There is a need also to differentiate the hearing loss caused by different occupational exposures that occurred at different times with different employers. Nearly all of these demands for allocation of hearing loss are in response to demands of the legal system. The legal profession is thus raising legitimate legal questions about hearing loss and about noise-induced hearing loss in particular.

ISO 1999 and Dobie Approach

An approach to the topic of allocation of hearing loss has been proposed by Dobie^{4,8,9} and reviewed. We do not wish to discuss the assumptions, rationale, strengths, or weaknesses of Dobie's approach. Rather, we wish to apply the method to some specific examples and examine the outcomes. In a second example, we wish to apply the ISO 1999¹ method and Dobie's method to a different set of data.⁶

Figures 17-7 to 17-10 are examples of the allocation of hearing loss to noise and to aging using the method described by Dobie (see Chapter 32). Very briefly, the validity of the Dobie method rests on the accuracy of the ISO 1999 standard including Data base A and B, on the assumption of additivity of noise-induced hearing loss and age-related hearing loss with

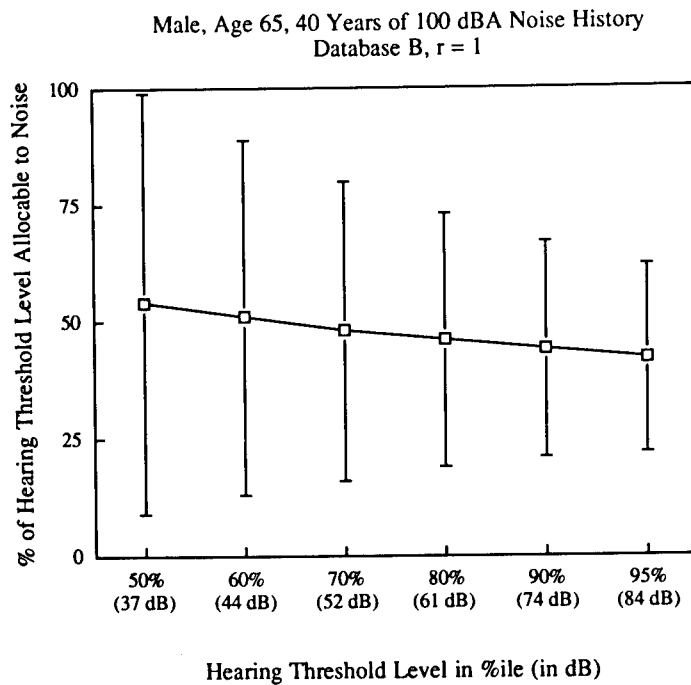
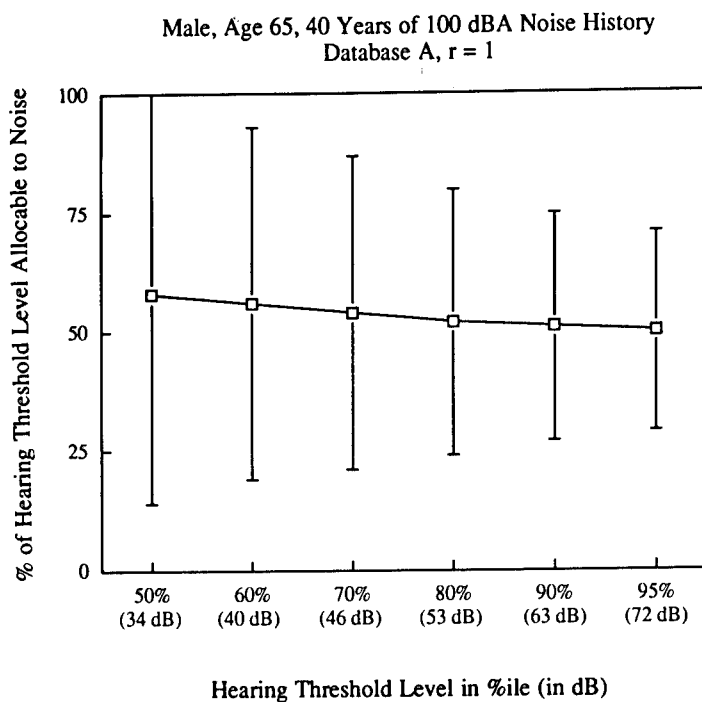
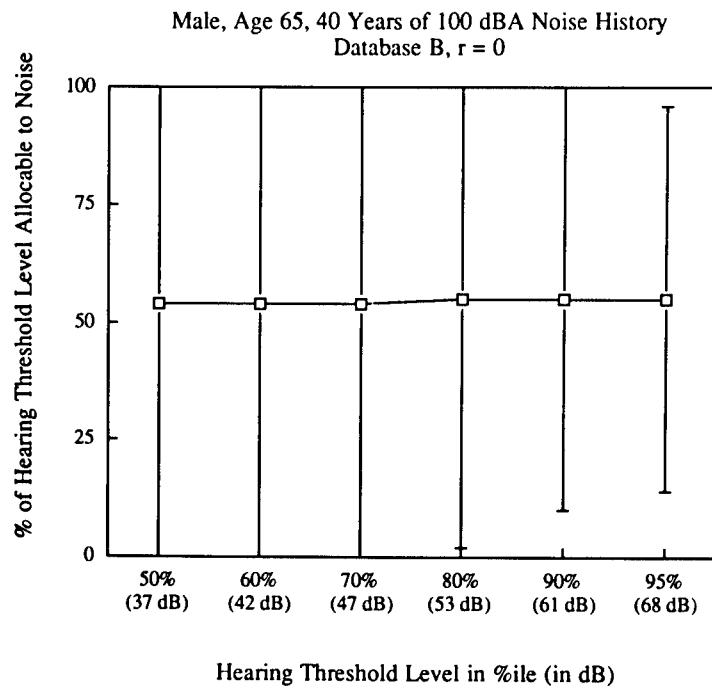
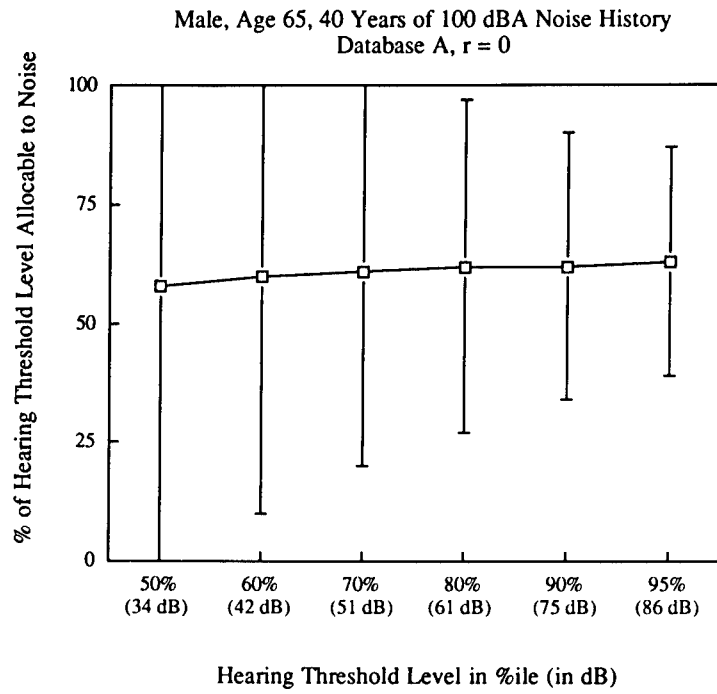


Figure 17-7 Examples of Dobie's allocation method for a male age 65 years, 40 years of exposure using ISO Data base A and B, and correlation coefficients of 1 or 0. Each panel shows the percentage of the hearing threshold level (mean 0.5, 1, 2, and 3 kHz) that is attributable to the noise exposure (100 dBA for 40 years at 8 h/d).



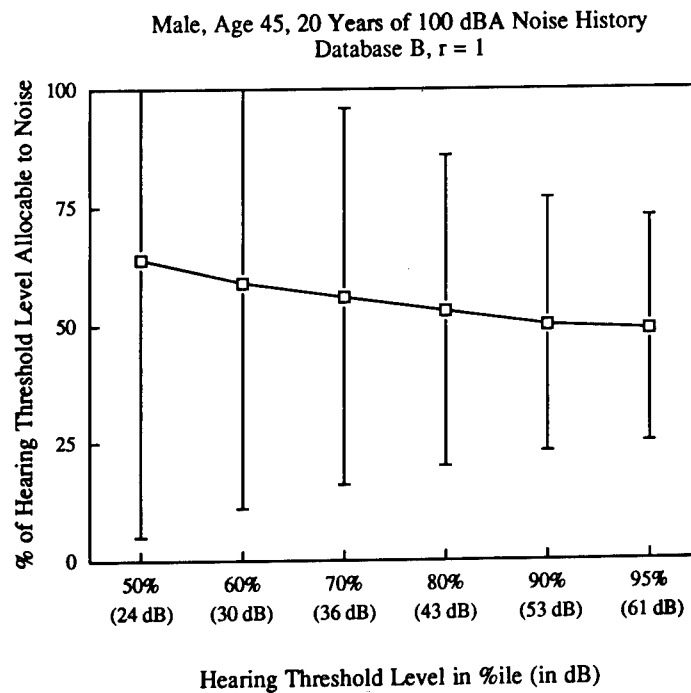
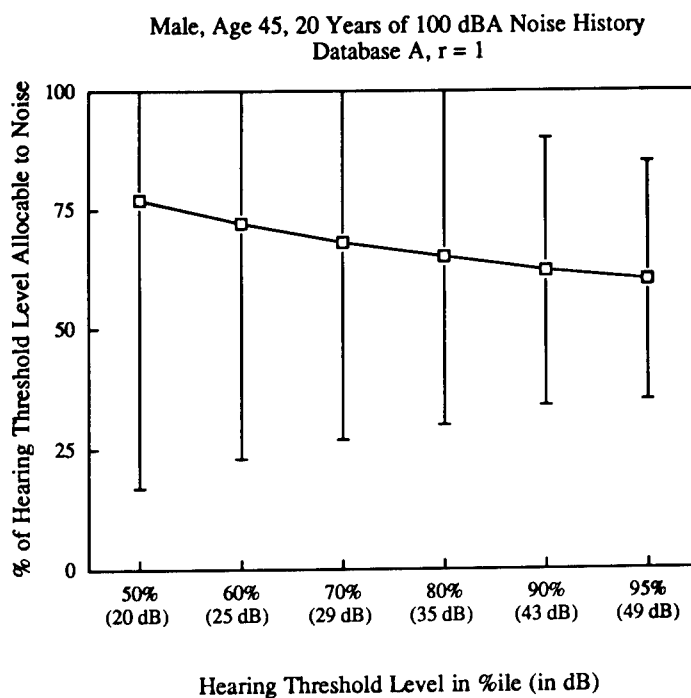
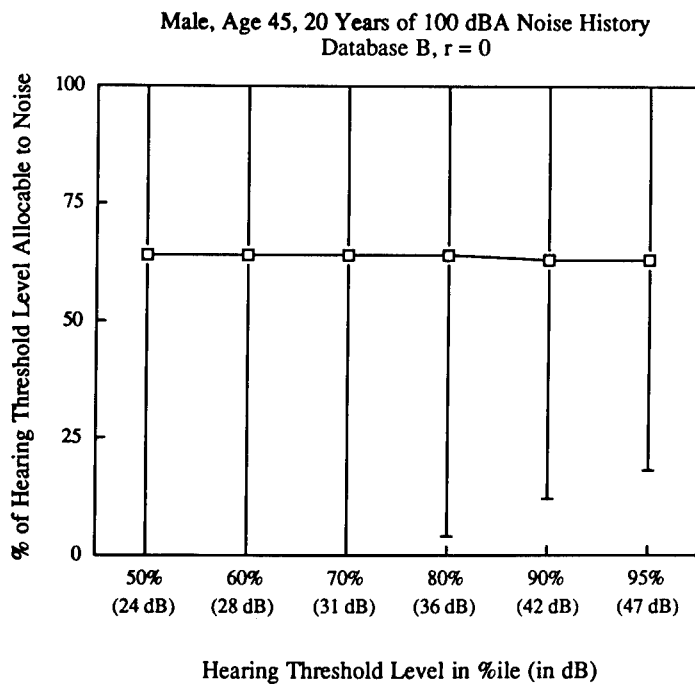
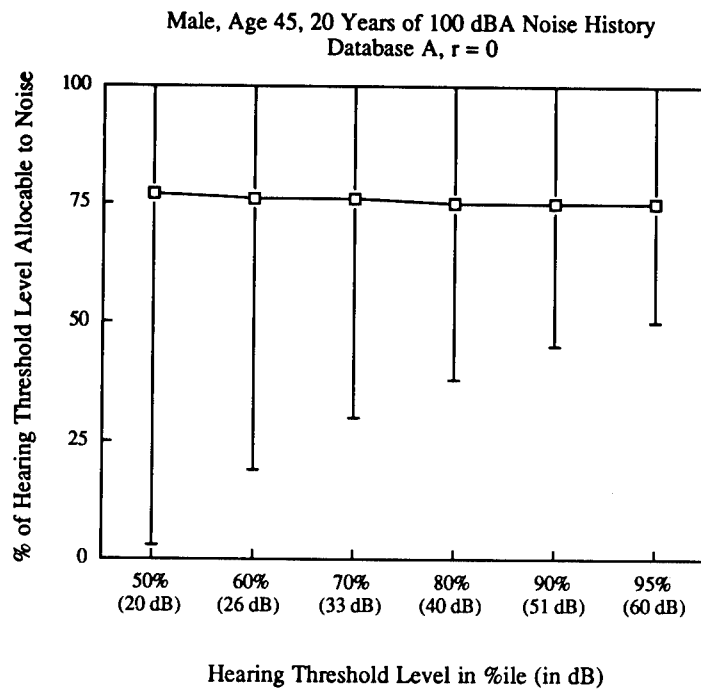


Figure 17-8 Data as in Figure 17-7 except the data are for a male age 45 years, 20 years of exposure at 100 dBA.



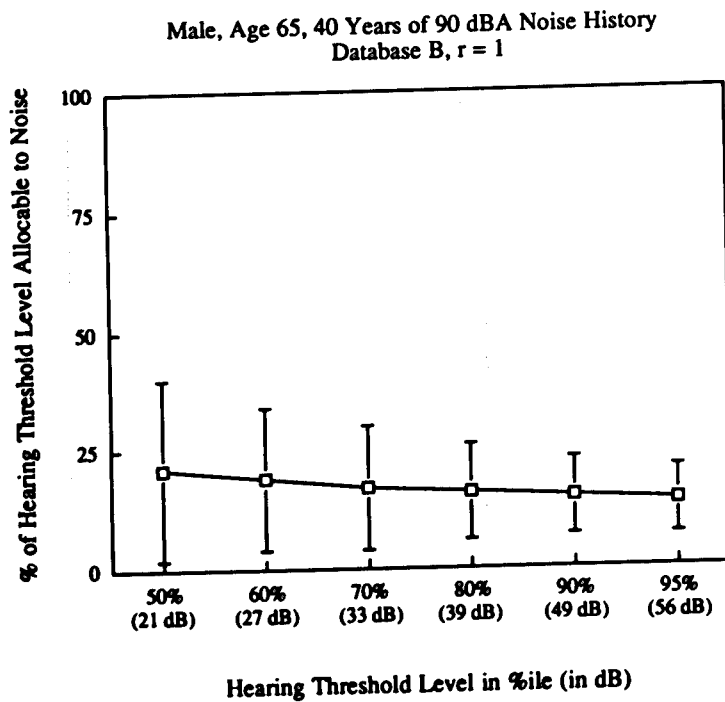
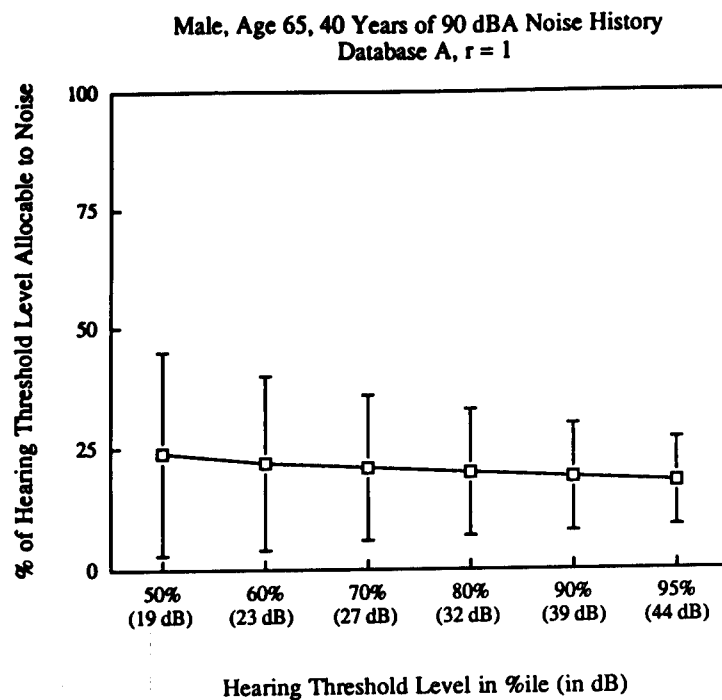
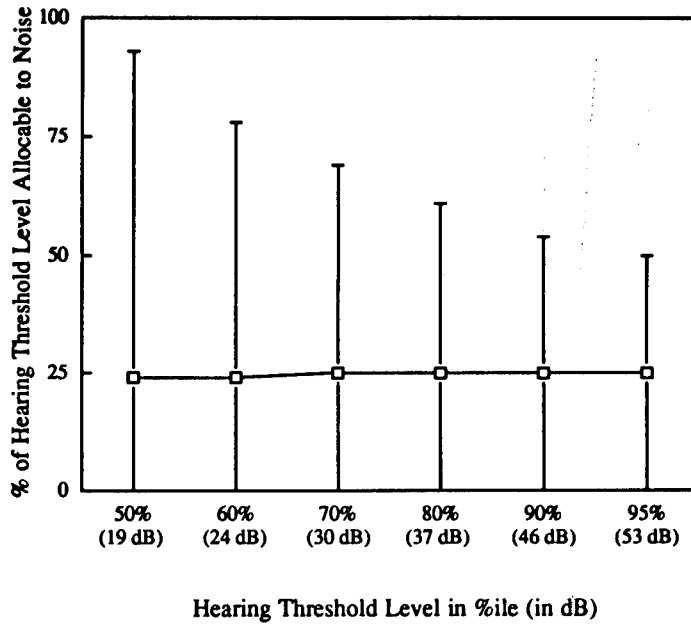
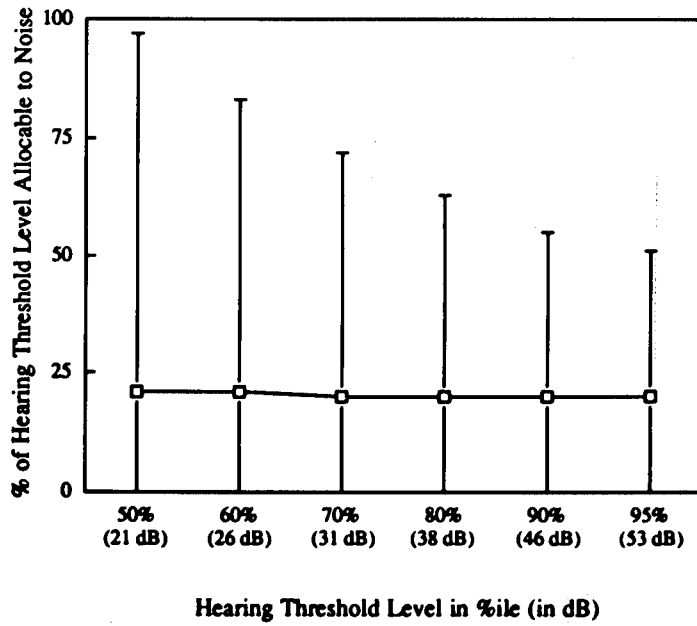


Figure 17-9 Data as in Figure 17-7 except the noise exposure level is 90 dBA rather than 100 dBA.

Male, Age 65, 40 Years of 90 dBA Noise History
Database A, $r = 0$



Male, Age 65, 40 Years of 90 dBA Noise History
Database B, $r = 0$



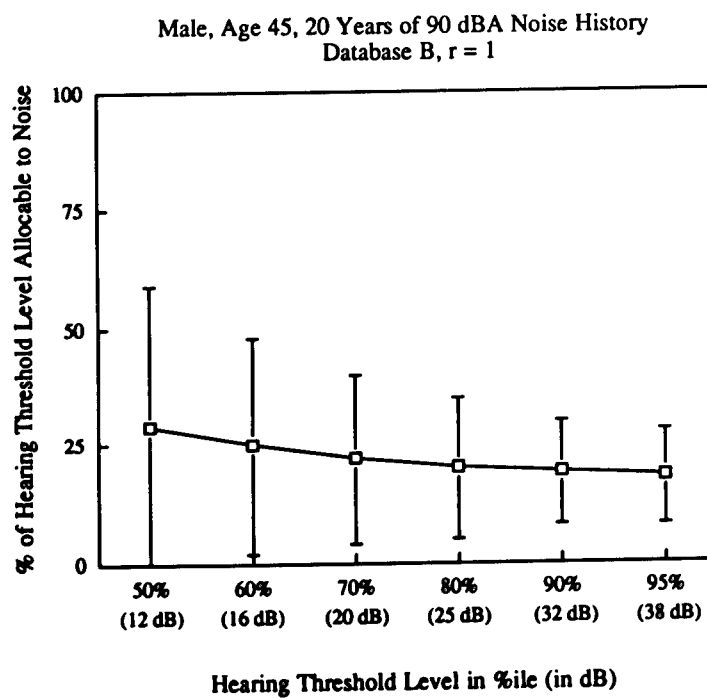
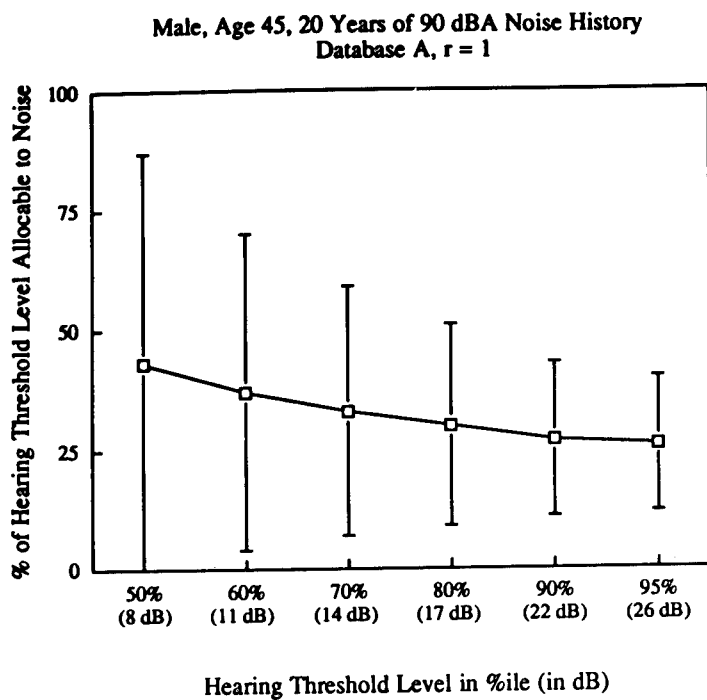
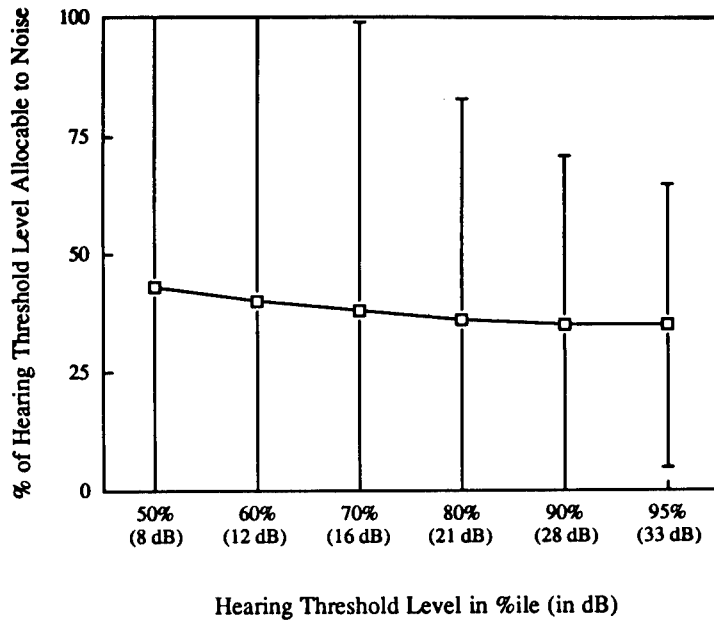
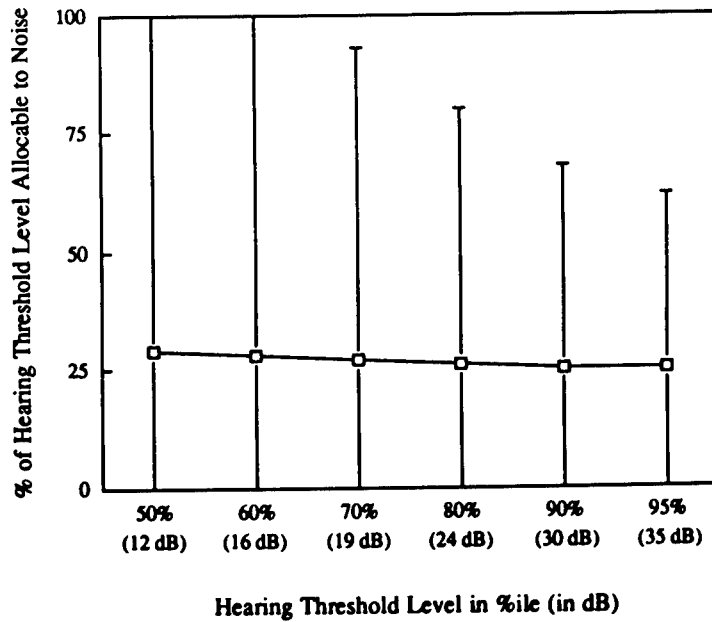


Figure 17-10 Data as in Figure 17-8 except the noise exposure level is 90 dBA.

Male, Age 45, 20 Years of 90 dBA Noise History
Database A, $r = 0$



Male, Age 45, 20 Years of 90 dBA Noise History
Database B, $r = 0$



a small compressive factor, and on Dobie's extension of the ISO 1999 standard (which was developed for and from large groups of persons) to individual persons. The accuracy of the model's predictions is improved by assuming a correlation between age-related and noise-induced hearing loss. There are other assumptions implicit and explicit in the Dobie model, but they are more procedural and second order than basic and conceptual, and are not discussed.

Figures 17-7 to 17-10 show the application of the Dobie method for different exposure conditions, and using correlation coefficients of $r = 0$ and $r = 1$. The ordinate on each figure is the percentage of the measured hearing loss attributable to noise exposure. Hearing loss is the mean of 0.5, 1, 2, and 3 kHz, which is the frequency grouping used and recommended by the American Academy of Otolaryngology, although any single frequency or grouping of frequencies can be used. The abscissa is the percentile of hearing loss starting at the 50th. Vertical bars indicate the 95% confidence interval.

Starting with Figure 17-7 and restricting the discussion to one or two points per figure, perhaps the most outstanding feature of Figure 17-7 across all four panels is that the percentage of the measured hearing loss allocatable to noise is about the same ($55 \pm 5\%$) for all subjects at age 65 with the same noise history. A second distinctive feature of Figure 17-7 is the magnitude of the 95% confidence interval. For $r = 1$ and hearing losses above the 70th percentile, the 95% confidence interval clearly fits within the probability space; for $r = 0$ using Data base B, the confidence interval does not fit within the probability space. The most striking feature in Figure 17-8 may be that the hearing loss allocated to noise clearly changes for $r = 1$ when comparing Data base A and B. It is also clear that aging plays less of a role for a male at age 45 than at age 65 (Figure 17-8), as one would obviously expect. For $r = 0$ and for many of the conditions where $r = 1$, the confidence interval exceeds the probability space.

When the noise level is 90 dBA as in Figures 17-9 and 17-10 rather than 100 dBA as in

Figures 17-7 and 17-8, it appears that about 20–25% of the measured loss is attributable to noise at age 65 years, regardless of all other variables. For $r = 1$ the confidence interval is contained within the probability space (Figure 17-9), whereas for all other conditions at 90 dBA at least some of the confidence intervals exceed the probability space.

In summary, two features of Figures 17-7 to 17-10 remain outstanding. One is that for 65-year-old males the percentage of the measured hearing loss attributed to noise remains the same regardless of the severity of the loss. The second outstanding feature is that for many of the conditions in Figures 17-7 to 17-10, the 95% confidence interval exceeds the probability space. Confidence intervals that exceed the probability space are particularly evident for conditions using Data base B and $r = 0$ between ARPTS and NIPTS.

With the publication of ISO 1999 and its application to many of the medical and legal aspects of NIPTS, it may be of interest to compare predictions made by ISO 1999 with actual epidemiological data, and to apply the ISO 1999–Dobie allocation method to the same epidemiological data. This is done in Figures 17-11 and 17-12 where the age-related threshold shift data are taken from Hinchcliffe.¹⁰ These control data were used by Taylor et al.⁶ in their classic field study of NIPTS. This study is considered classic because it is one of the few where the occupational noise probably remained constant over a period longer than 40 years. In most field studies, one is never certain of the changes that have occurred in the noise characteristics.

Figure 17-11 (top half) compares the Hinchcliffe normative aging data for women¹⁰ with predictions from ISO 1999.¹ The Hinchcliffe data are the normative data used in the Taylor et al.⁶ study of NIPTS in jute weavers. The bottom half of Figure 17-11 compares the Taylor et al.⁶ NIPTS data with predictions from ISO 1999.¹ Perhaps the most outstanding feature of the top half of this figure is the close correspondence between the Hinchcliffe aging data at the 50th percentile and the ISO Data base A, 50th percentile. The best fit for the Hinchcliffe data¹⁰ was a second-order

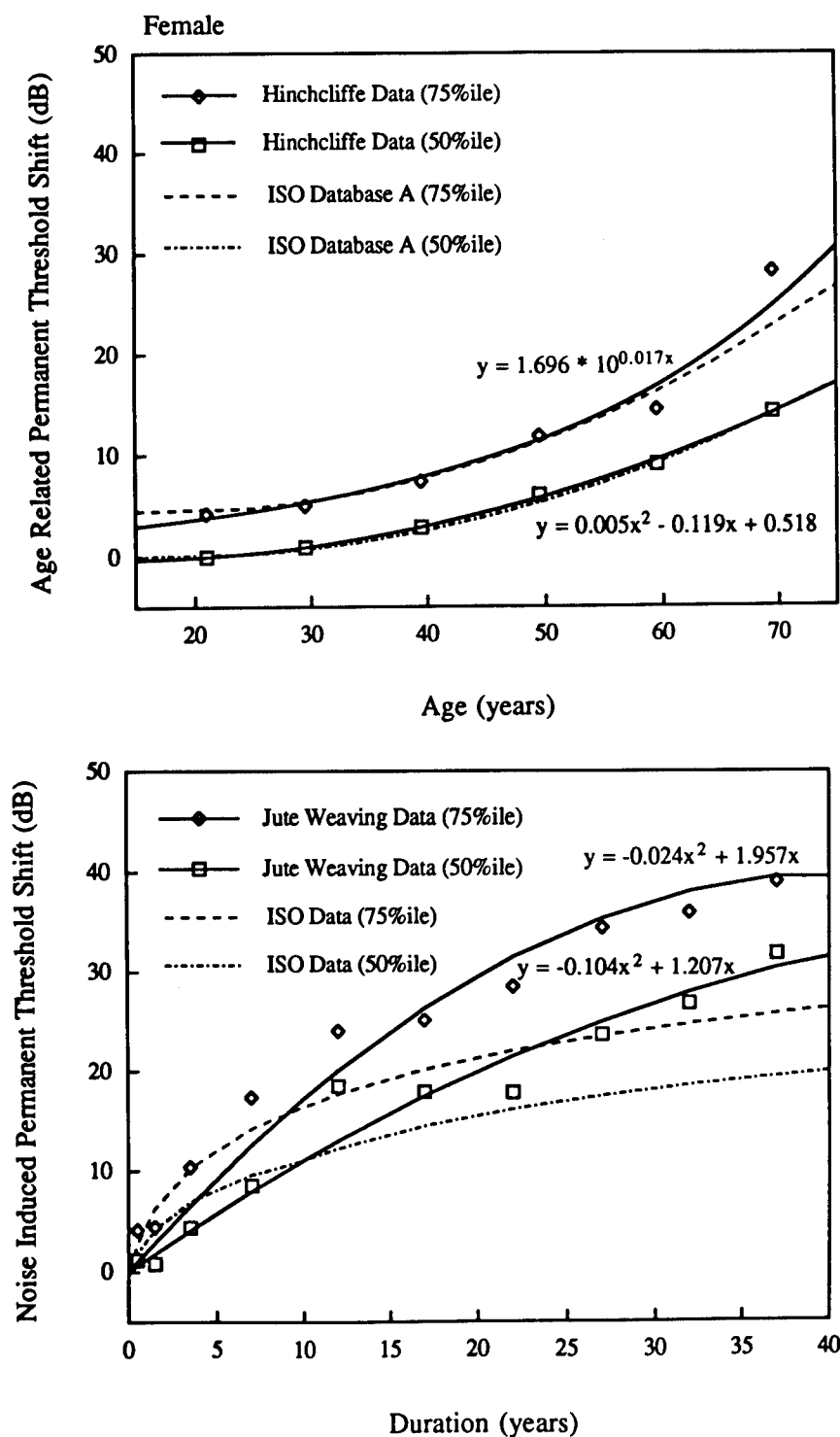


Figure 17-11 Comparison of ISO 1999¹ with the Hinchcliffe aging data¹⁰ and the jute weaving data of Taylor et al.⁶ The top panel compares the 50th and 75th percentile data of Hinchcliffe with the 50th and 75th percentile of ISO Data base A. In the bottom panel NIPTS data predicted by ISO 1999 for the Taylor et al. data are compared to the Taylor et al. data. The equations are the result of the application of ISO 1999 methods to the Taylor et al. data.

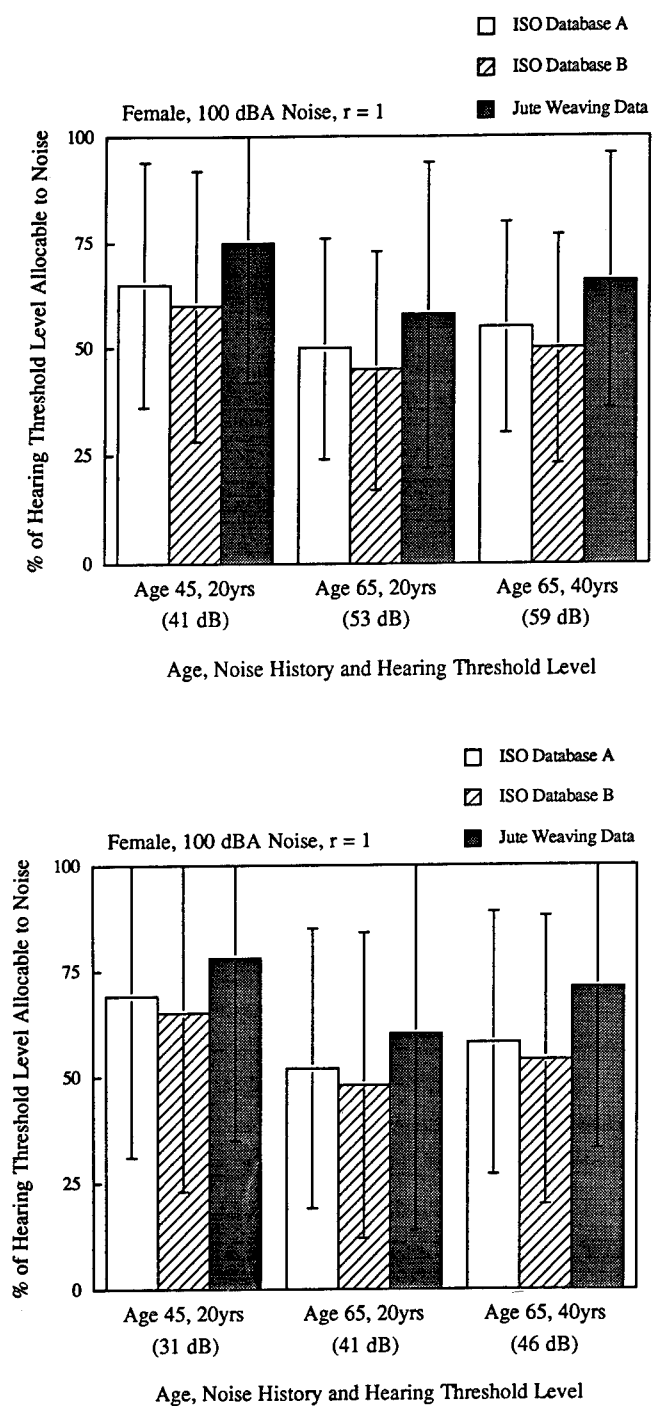
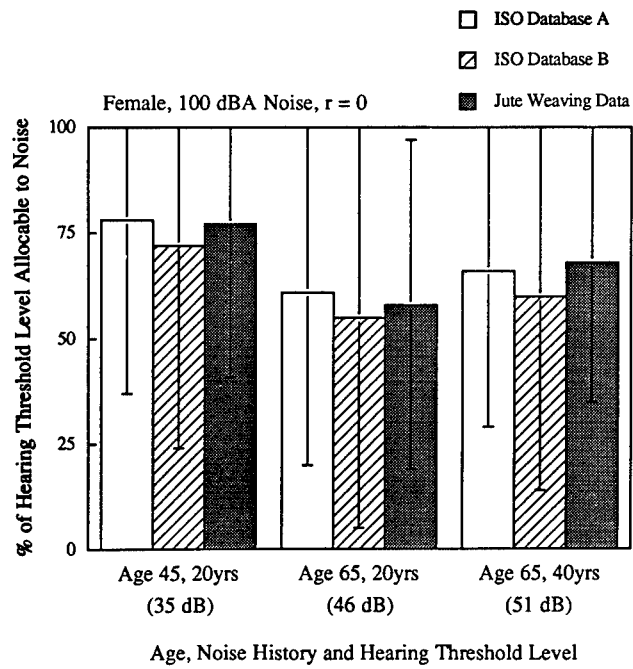
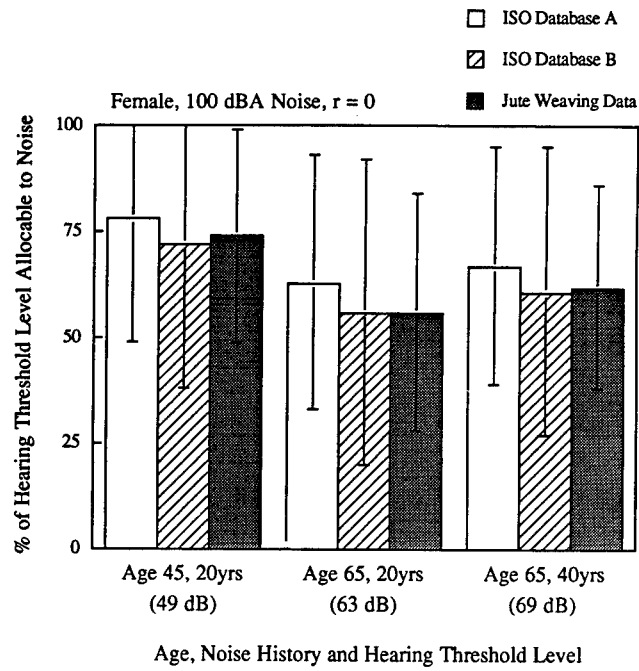


Figure 17-12 The Dobie allocation method^{4,8,9} has been applied to specific conditions of the Taylor et al.⁶ data from Figure 17-11. The top panel is the 90th percentile data and the bottom panel is the 75th percentile.



polynomial for the 50th percentile and an exponential equation for the 75th percentile. Logarithmic models of ISO 1999¹ did not fit the data as accurately. Likewise, the NIPTS data from Taylor et al.⁶ were described best by second-order polynomials rather than the logarithmic functions of ISO 1999.¹

Perhaps the most outstanding feature of the bottom half of Figure 17-11 is the close correspondence between the ISO predictions and the Taylor et al. data in the first 10 years of exposure and the wide disparity after 25 and 30 years of exposure.^{1,6} We have no explanation for the close agreement at 10 years of exposure and the poor agreement at greater than about 25 years.

Figure 17-12 shows the application of the Dobie method to three hypothetical female subjects, age 45, with 20 years of exposure; age 65, with 20 years of exposure; and age 65, with 40 years of exposure. Both the ISO 1999 data base and the Hinchcliffe-Taylor et al. data are used. The top half of Figure 17-12 assigns our hypothetical female subjects to the 90th percentile and the bottom half assigns them to the 75th percentile. Note in Figure 17-12 as in the previous figures, the 95% confidence interval is given by the vertical lines. The differences resulting from using Data base A or B were never greater than 4–6%. Predictions using the Hinchcliffe-Taylor et al. data were 8–13% higher than those predictions derived from ISO using Data base A. It is noteworthy that differences of 8–13% are not statistically significant at the usual and customary probability of 0.05.

Summary

Data from a group of laboratory animals (gerbils) who were born and reared in a quiet environment and from a group who were exposed to noise for about 2 years, support the concept of additivity of NIPTS and age-related threshold shift. On an individual animal basis, the variability in age-related threshold shift under highly controlled conditions is very large with hearing losses ranging from 0 to >70 dB. Such individual variability in control animals necessarily complicates the allocation of hearing

loss in individuals to an age-related component and a noise-related component.

Epidemiologic data from human subjects also show considerable variability that is dependent on duration of the exposure, level of the exposure, and test frequency. Application of ISO 1999¹ and the Dobie methods^{4,8,9} provide very specific quantitative assessments of noise and aging components; however, the 95% confidence interval for a given allocation in a given individual often covers the entire probability space.

Acknowledgment

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Chapter 18

Application of Frequency and Time Domain Kurtosis to Assessment of Complex, Time-Varying Noise Exposures

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Noise measurements made for the purpose of hearing conservation practice have as their objective the extraction of some physical metric from the noise that can be used to estimate the hazards to hearing from prolonged exposure to that noise. In many industrial environments, the conventional variables used to characterize noise exposure such as sound pressure level (SPL) and spectrum vary considerably from moment to moment over an 8 hour work day. The noise typically consists of a time-varying continuous background noise with varying superimposed impulsive or other transient components.¹ Despite similar sound pressures and spectra, complex noises can vary appreciably in their statistical properties and can produce different amounts and frequency distributions of hearing loss in exposed individuals.

Demographic data²⁻⁵ have shown that non-Gaussian noise exposures are more hazardous to hearing than are Gaussian noises of similar L_{eq} . Current practice relies primarily on measures such as the A-weighted L_{eq} , a metric that completely ignores the temporal characteristics of an exposure. The role of temporal variables has been most recently emphasized in studies using interrupted noise exposure paradigms or preexposure priming noises,^{6,7} both of which produce a resistance to developing threshold shifts (TSs) from latter exposures.

Background Data

An early experiment by Hamernik et al.⁸ suggested that complex noises (i.e., combinations of Gaussian noise and impacts) had the potential to cause greater damage than would be anticipated based upon the effects that either of the two classes of noise presented alone would produce. They combined a 95 dB octave band of noise centered at 0.5 kHz with brief 158 dB impulses (A duration⁹ \approx 40 microseconds) presented 1/min for 1 hour. The combination exposure produced significantly more permanent TS (PTS) and sensory cell losses than either of the impulse or continuous noises components. The PTS for these three exposures are shown in Figure 18-1. Furthermore, when the impulse, which had a predominately high frequency (>4.0 kHz) energy content, was presented at a rate of 1/min during a 2 second quiet interval in the 95 dB SPL octave band of noise, the interaction effect disappeared. This series of exposures suggested that the two classes of noise could interact to exacerbate hearing loss to an extent that would not be anticipated based upon energy considerations alone and that temporal variables of an exposure paradigm could modulate this interaction effect.

Since that time a number of experiments have been carried out using complex noise exposure paradigms¹⁰⁻¹³ that demonstrated

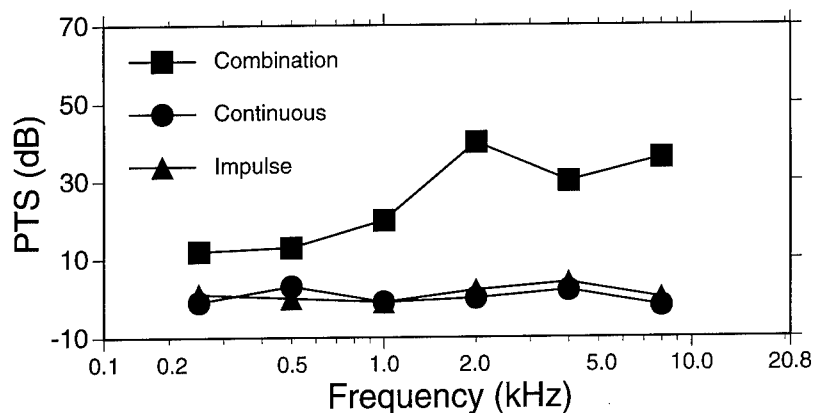


Figure 18-1 The mean permanent threshold shift (PTS) audiogram for the groups exposed to either the 158 dB peak SPL impulse noise, the 0.5 kHz, 95 dB SPL octave band continuous noise, or the combination of these two exposures.

the problems associated with the use of time-averaged (energy) metrics to estimate the hazards of an exposure. A summary of the results from several experiments using complex noise exposures was prepared by Ahroon et al.¹⁴ and is illustrated in Figure 18-2. Details of the experimental paradigms and an explanation of the symbols used in this figure can be found

in Ahroon et al.¹⁴ This figure illustrates the mean total outer hair cell (OHC) loss for groups of animals exposed to various complex noises having sound exposure levels (SELs) in the range 140–160 dB {where $SEL = 10 \log_{10} \int_0^t [P^2(t)/P_o^2] dt$, and $P_o = 20 \mu Pa$ and $t_o = 1$ second}. In a number of the cases shown, exposures having similar SELs produced widely

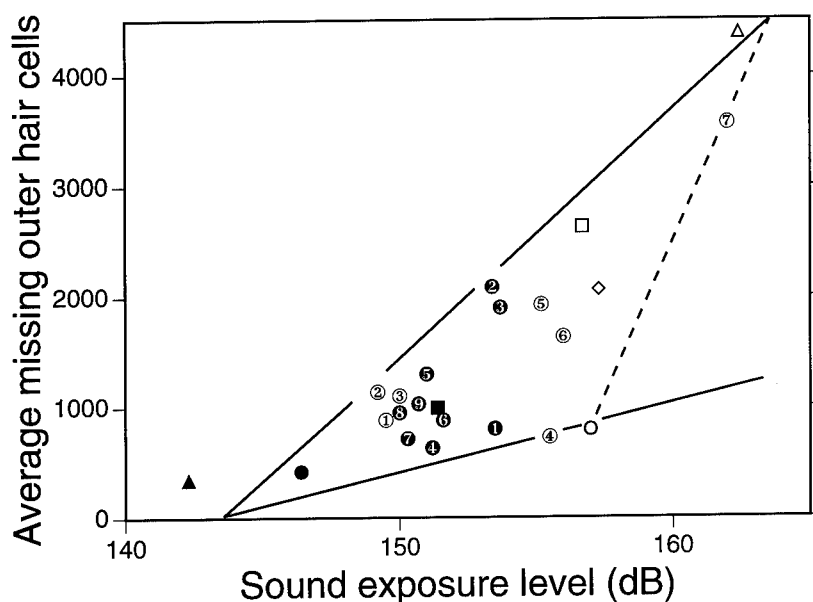


Figure 18-2 The mean total number of outer hair cells lost in groups of animals exposed to noises having the indicated sound exposure levels. Symbols are explained in Ahroon et al.¹⁴

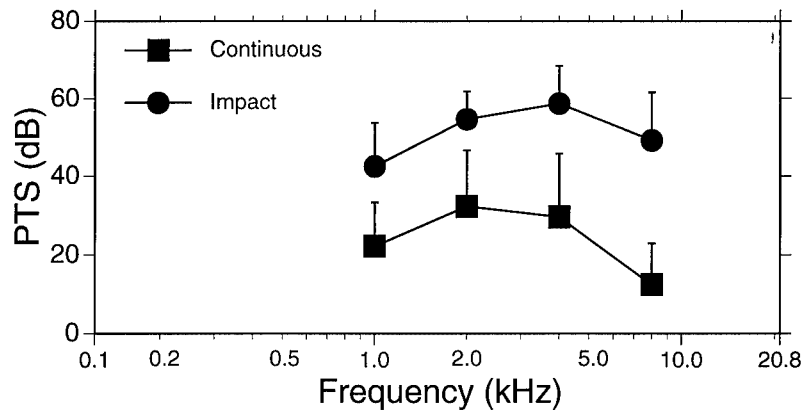


Figure 18-3 The mean permanent threshold shift (PTS) audiograms from exposure to either continuous or impact noises having the same total energy and spectra.

differing sensory cell pathology, indicating that metrics other than energy indices are needed in order to identify and quantify hazardous exposures. The exacerbation of hearing loss following exposure to complex non-Gaussian noise environments is not too surprising when one considers differences in the mode of damage production between high-level noise exposure and more modest levels¹⁵⁻¹⁷ and the differences in trauma produced by continuous noise and noise transients having equal energy and spectra. For example, Figure 18-3, replotted from Dunn et al.,¹⁸ shows up to 30 dB greater PTS from impulsive noise exposure than from an energy and spectrally equivalent continuous noise. The levels used were 136 dB peak SPL and 110 dB SPL, respectively, with the duration of exposure balanced to produced equal-energy exposures.

The above results collectively show that: there are differences in the extent of trauma produced by energy and spectrally equivalent impact and continuous noise; that the two classes of noise can interact in unexpected ways to exacerbate or otherwise modify acoustic trauma; and that the temporal characteristics of an exposure can affect the extent of trauma. Such conclusions suggest the need to develop a metric to gauge the hazards of an exposure that takes into account the temporal

and peak distribution properties of a noise exposure.

Experimental Methods

The chinchilla was used as the animal model in each of the experiments described. All indices of hearing were obtained using the auditory-evoked potential recorded from chronic electrodes implanted in the inferior colliculus and all quantitative histological evaluations relied upon surface preparations of the organ of Corti. Complete details of the experiments that will be described can be found in the indicated references.

The noise exposures were generated using digital technology. Each exposure was designed in the frequency domain by manipulating the phase spectrum and applying the inverse Fourier transform to create the time domain signal. The noise generation system allowed control over the non-Gaussian structure of the temporal signal. Families of equal-energy and equivalent long-term spectra noises could be generated whose kurtosis statistic could be varied through probability functions that controlled the occurrence and intensity of transient components. Furthermore, the regions of the spectrum that contributed to the formation of the transients was under ex-

perimental control. Details of the system are described in Hseuh and Hamernik.^{19,20}

Recent Results

A series of experiments²¹ was designed to determine if, for equal L_{eq} and power spectra, the effects on hearing of high temporal kurtosis, $\beta(t)$, noise exposures, and a Gaussian noise exposure were different. $\beta(t)$ was computed over a 205 millisecond window and an average value obtained over 5000 samples. The three noise exposures consisted of a shaped, broadband noise whose spectrum is shown in Figure 18-4. Each noise was presented at the relatively low level of 90 dB SPL and lasted continuously for 5 days. The impulsiveness defined by the $\beta(t)$ and the region of the spectrum from which the impulsive components of the noise were created differed for two of the noises, while the third was a continuous Gaussian noise. Impact peak levels in the two non-Gaussian continuous noise exposures were set at 117 dB ($\beta(t) = 189$) or 114 dB ($\beta(t) = 27$). Three groups of chinchillas (10 per group) were exposed to one of the three exposures. The audiometric results showed that the high-kurtosis exposure produced up to 20 dB greater hearing loss at the high frequencies

and resulted in differences in the pattern of sensory cell loss. The pattern of sensory cell loss could not be reconciled with the frequency profile of the audiometric loss.

The PTS for the three groups is shown in Figure 18-5a. Immediately apparent are the clear differences in PTS at the high frequencies (at and above 4.0 kHz) between the $\beta(t) = 189$ and the other two exposure conditions. A mixed-model analysis of variance confirmed a statistically significant main effect of kurtosis ($F = 6.46$, $df = 2/27$, $p < 0.01$) and a statistically significant interaction between kurtosis and frequency ($F = 2.86$, $df = 12/162$, $p < 0.01$). As seen in the figure, PTS across groups was very similar at 0.5, 1.0, and 2.0 kHz, but the three groups diverged at the higher frequencies with the Gaussian exposure producing the least PTS. The main effect of frequency was also significant. A separate mixed model ANOVA was performed comparing the PTS from the $\beta(t) = 3$ and $\beta(t) = 27$ exposure groups. The results indicated a statistically significant main effect of kurtosis ($F = 5.51$, $df = 1/18$, $p < 0.05$), a significant main effect of frequency ($F = 3.76$, $df = 6/108$, $p < 0.01$), and a significant interaction between kurtosis and frequency ($F = 3.14$, $df = 6/108$, $p < 0.01$). The $\beta(t) = 27$ exposure resulted in significantly

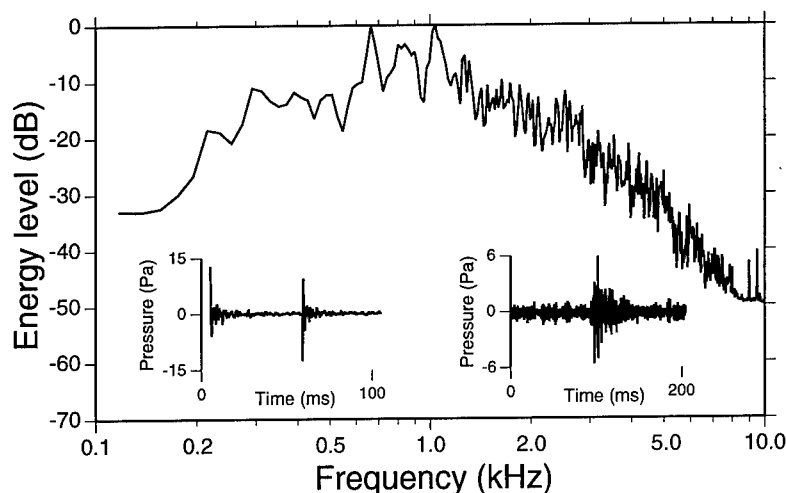


Figure 18-4 The average relative energy spectrum of the 90 dB SPL continuous noise exposures. The pressure-time waveforms illustrate the shape of the 117 and 114 dB peak SPL impacts that were present in the non-Gaussian continuous noise conditions.

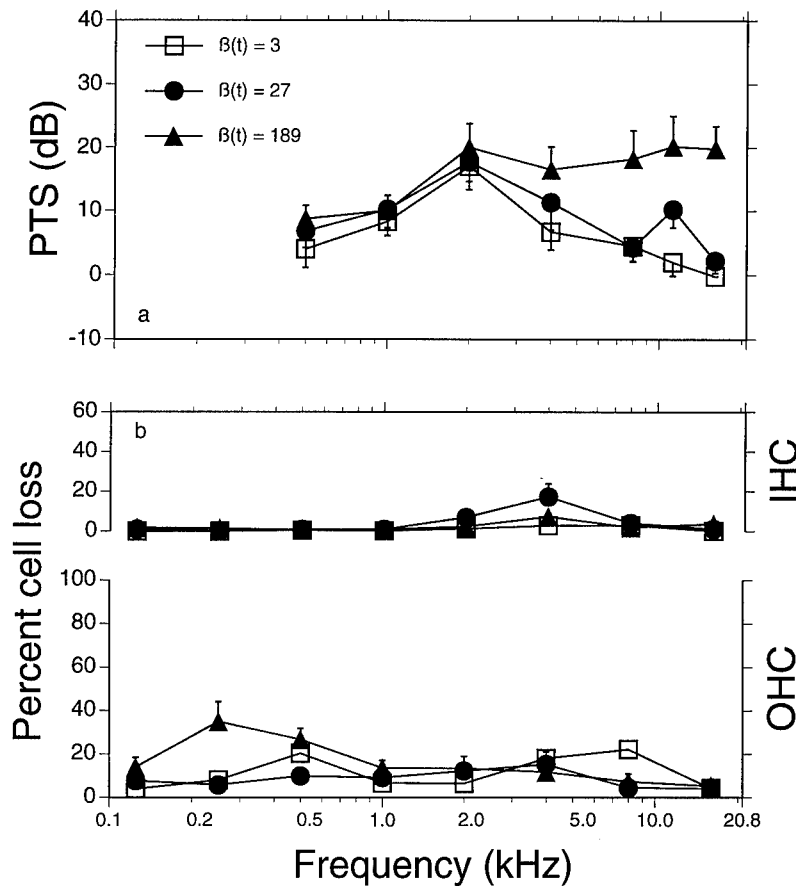


Figure 18-5 (a) Mean permanent threshold shift (PTS) audiograms and (b) inner and outer hair cell losses (IHC, OHC) for groups of animals exposed to noises having the indicated time domain kurtosis, $\beta(t)$.

more PTS than the Gaussian exposure [$\beta(t) = 3$] only at 11.2 kHz, amounting to an 8.2 dB effect ($t = 2.361$, $df = 18$, $p < 0.05$).

The group mean inner and outer hair cell (IHC, OHC) loss are shown in Figure 18-5b. A mixed-model analysis of variance on the percent sensory cell losses indicated that the main effect of frequency was statistically significant. However, for both IHC and OHC losses, the main effects of kurtosis were not significant (IHC: $F = 1.36$, $df = 2/27$; OHC: $F = 2.33$, $df = 2/27$), but both interactions between kurtosis and frequency were statistically significant (IHC: $F = 2.61$, $df = 14/189$, $p < 0.01$; OHC: $F = 3.36$, $df = 14/189$, $p < 0.01$). The significant interaction for the percent IHC-dependent measure appears as elevated losses in the two higher kurtosis groups at the location of the

basilar membrane associated with the 4.0 kHz test frequency; and the percent OHC interaction is the result of elevated losses at the locations corresponding to 0.25 kHz for the $\beta(t) = 189$ group and at 0.5 kHz in the $\beta(t) = 3$ and $\beta(t) = 189$ groups. That is, there were no significant differences in the mean percent sensory cell loss across the three groups, but there were significant differences in the distribution of sensory cell loss across frequency (i.e., place). To gain a different perspective on the distribution of sensory cell losses across the three groups of noise-exposed animals, the 10 individual cochleograms in each group representing IHC and OHC losses are plotted in Figures 18-6 and 18-7, respectively. In these figures, the 10 cochleograms for each group have been superimposed, and rather than be-

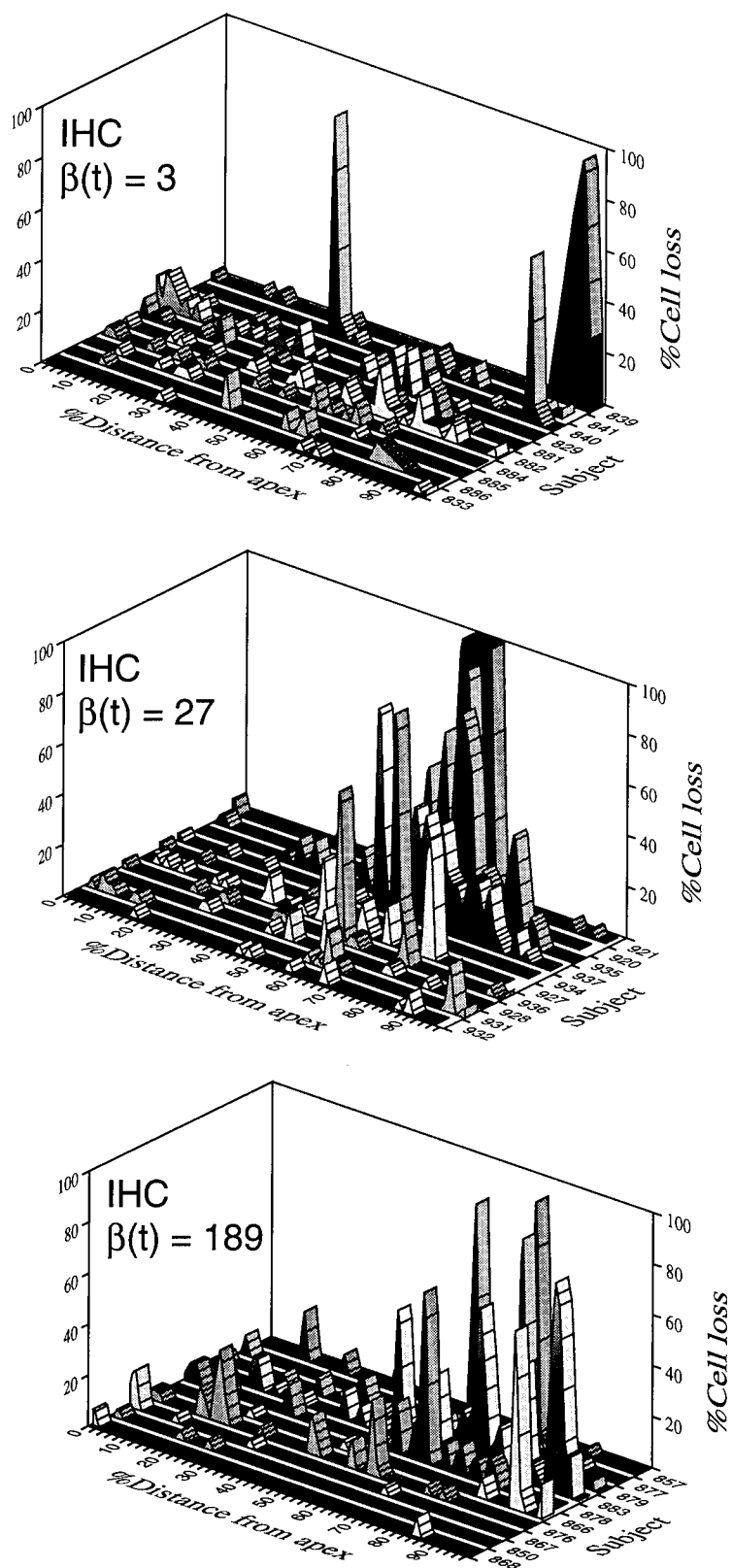


Figure 18-6 A superposition of all 10 individual animal inner hair cell (IHC) cochleograms following exposure to the noises having the indicated time domain kurtosis, $\beta(t)$.

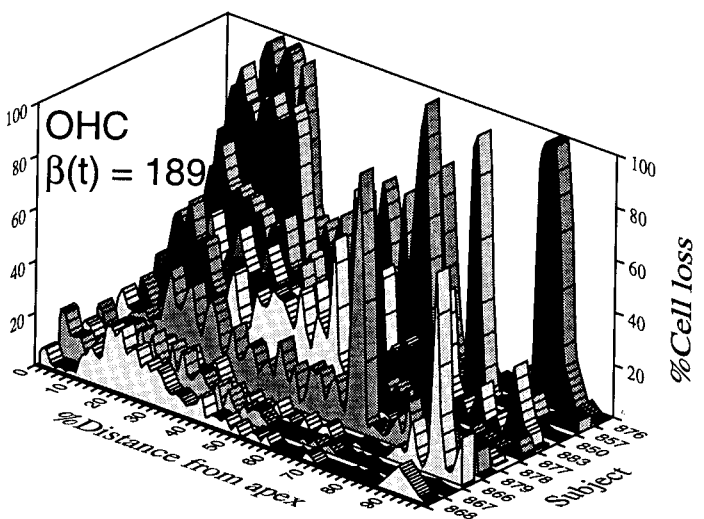
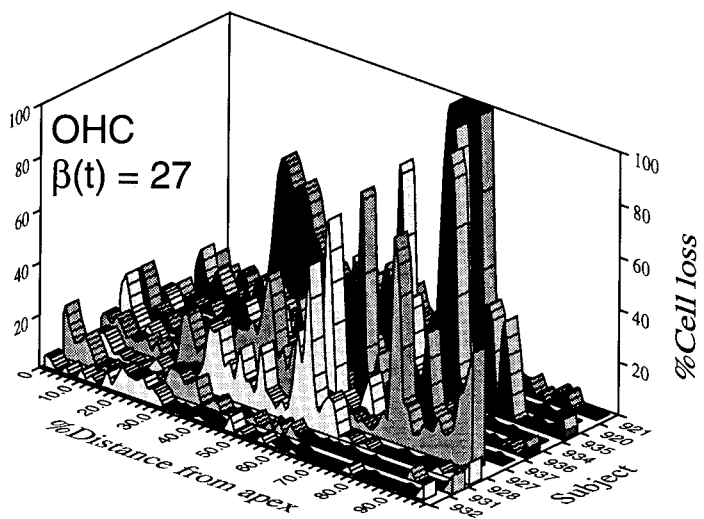
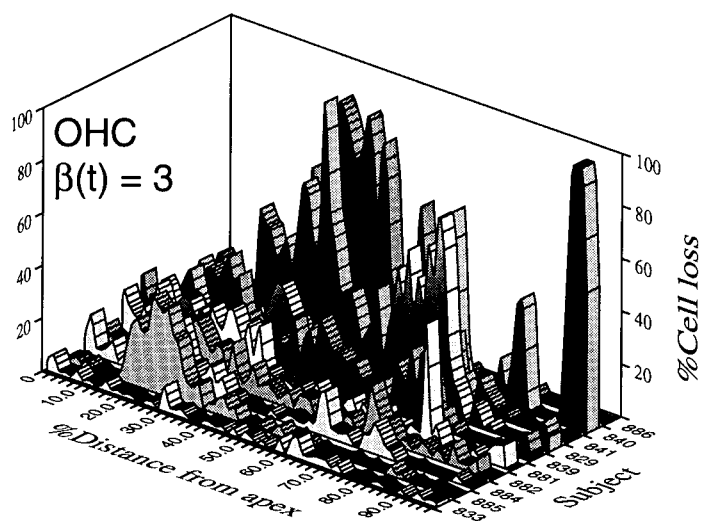


Figure 18-7 A superposition of all 10 individual animal outer hair cell (OHC) cochleograms following exposure to the noises having the indicated time domain kurtosis, $\beta(t)$.

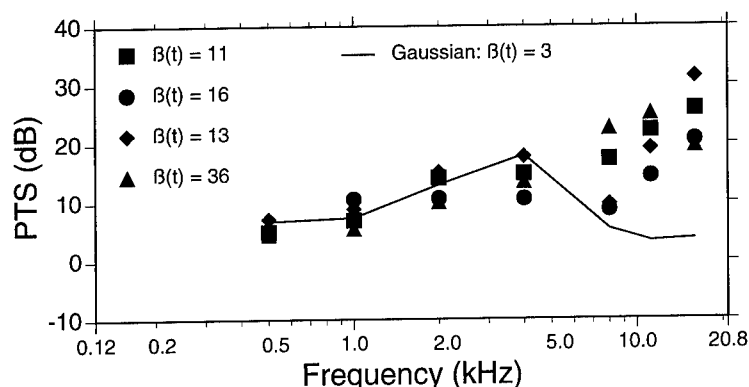


Figure 18-8 The permanent threshold shift (PTS) audiograms for the five equal-energy (97 dB SPL) noise exposures having the indicated time domain kurtosis, $\beta(t)$.

ing able to read the details of any given animal, only an impression of the variability in the distribution of cell loss on a group basis is discernible.

Related Experiment

Another series of equal-energy and spectra exposures consisting of four complex noises and one Gaussian noise exposure was conducted using the same noise spectrum as in the previously described exposures (Figure 18-4). Each exposure that lasted for five days was presented at an L_{eq} of 97 ± 2 dB SPL. The four complex noises were a mixture of Gaussian noise and synthesized impacts having the $\beta(t)$ values shown in Figure 18-8. The impacts, however, were presented with varying probabilities of occurrence ($1/10$ or $1/50$) and at peak SPL levels that were either 118 dB for the $1/10$ probability group or 123 dB for the $1/50$ probability group. The peak levels were fixed for a given exposure. The audiometric results for these five groups are shown in Figure 18-8. As with the audiometric results from the previous experimental series, there were again statistically significant differences in PTS between the Gaussian and non-Gaussian noises at the highest test frequencies. Similar high-frequency differences were seen in the data obtained from the human population studied by Fausti et al.²²

Thus, the two sets of experimental data presented above showed that at various L_{eq} 's,

there were differences in the audiometric results at the high frequencies for classes of exposures having the same spectra and L_{eq} . Notice that the high-frequency loss in the second experiment is less than that in the first experiment despite a higher L_{eq} . The impacts used in the exposures of the first experiment took place once every 204 milliseconds and had peaks of 117 dB. The impacts also had a different temporal structure with an overall duration of roughly 5 milliseconds. The impacts in the complex noise of the second set of experiments occurred randomly and were, on average, either 10 or 50 times less frequent. They also had a different temporal structure and had an overall duration of 50 milliseconds. Thus, while both data sets showed that complex noises having the same spectrum produced greater trauma at the high frequencies than did an equivalent L_{eq} Gaussian noise, the results suggested that the magnitude of the loss might be related to temporal factors in the impact presentation schedule. Note also the differences in $\beta(t)$ in each of these two series of experiments.

Application of Frequency Domain Kurtosis to Assessment of Non-Gaussian Noise Exposures

Five different noise exposures were used in this experiment.^{23,24} All had an L_{eq} of 100 dB and the same, reasonably uniform (flat) spectrum from 0.2 to 10 kHz. The exposures lasted

continuously for 5 days. The exposures are identified as follows:

- CN V: Broadband (0.2–10 kHz) Gaussian noise having $\beta(t) \approx 3$. The spectrum of this noise is shown in Figure 18-9a along with a 50 millisecond sample of the temporal wave form (inset).
- CN VI: A combination of Gaussian noise and impacts having the average spectrum shown in Figure 18-9a and $\beta(t) = 84$. An example of the pressure–time history of the impact and its spectrum is shown in Figure 18-9b. The impact peak SPL was 125 dB and the impacts had a probability of occurring in a 50 millisecond window of 0.1. The spectrum of the Gaussian noise that was combined with the impacts to produce the high kurtosis complex noise is shown in Figure 18-9d as the solid line.
- CN VII: This complex noise was similar to CN VI except that instead of impacts, a noise burst having a duration of 50 milliseconds and an rms level of 106 dB SPL was presented embedded in the Gaussian background noise. The probability of a burst occurring in a 50 millisecond window was set at 0.1. The combination of a transient noise burst and a continuous background noise had $\beta(t) = 21$. The overall average spectrum of this complex noise is the same as that shown in Figure 18-9a; the spectrum and wave form of the noise burst is shown in Figure 18-9b, and the spectrum of the background noise is shown in Figure 18-9d as the solid line.

Thus noises CN V, CN VI, and CN VII had similar overall spectra and were presented at the same L_{eq} , but differed in their impulsiveness defined by the temporal kurtosis, $\beta(t)$.

- CN VIII: This noise was a combination of Gaussian noise and impacts having the average spectrum shown in Fig-

ure 18-9a and $\beta(t) = 45$. The impacts had the spectrum and time history shown in Figure 18-9c. The impacts whose energy spectrum peaked in the 1.0–4.0 kHz region had a 126 dB peak SPL and a 0.1 probability of occurring in a 50 millisecond window. The spectrum of the Gaussian noise that was combined with the impacts is shown in Figure 18-9d as the dotted line from 0.1 to 1.5 kHz and then as the solid line from 1.5 through 10.0 kHz.

- CN IX: This complex noise was similar to CN VIII except that instead of impacts, a noise burst having a 50 millisecond duration and an rms level of 104 dB SPL was presented. The probability of a burst occurring in a 50 millisecond window was set at 0.1. This complex combination of continuous noise and noise bursts had a $\beta(t) = 10$. The average overall spectrum of the complex noise is shown in Figure 18-9a; the spectrum and wave form of the noise burst is shown in Figure 18-9c and the spectrum of the Gaussian background noise is shown in Figure 18-9d as the dotted line for the low frequency portion (to 1.5 kHz) and the solid line for frequencies greater than 1.5 kHz.

Thus noises CN V, CN VIII, and CN IX each had a similar overall spectrum, were presented at the same L_{eq} , but differed in their impulsiveness defined by $\beta(t)$. Also evident from the spectra shown in Figure 18-9 is that the exposure pair CN VI and CN VII and pair CN VIII and CN IX, while having the same energy and spectra, differed in the regions of the spectrum from which the higher energy transients (whether noise bursts or impacts) derived their energy.

The design and digital generation of the acoustic signal is detailed in Hsueh and Hamernik.^{19,20} The acoustic signal was sampled at 20 kHz with 1024 samples per 51.2 millisecond window. The time domain kurtosis, $\beta(t)$, was computed over approximately a 256 sec-

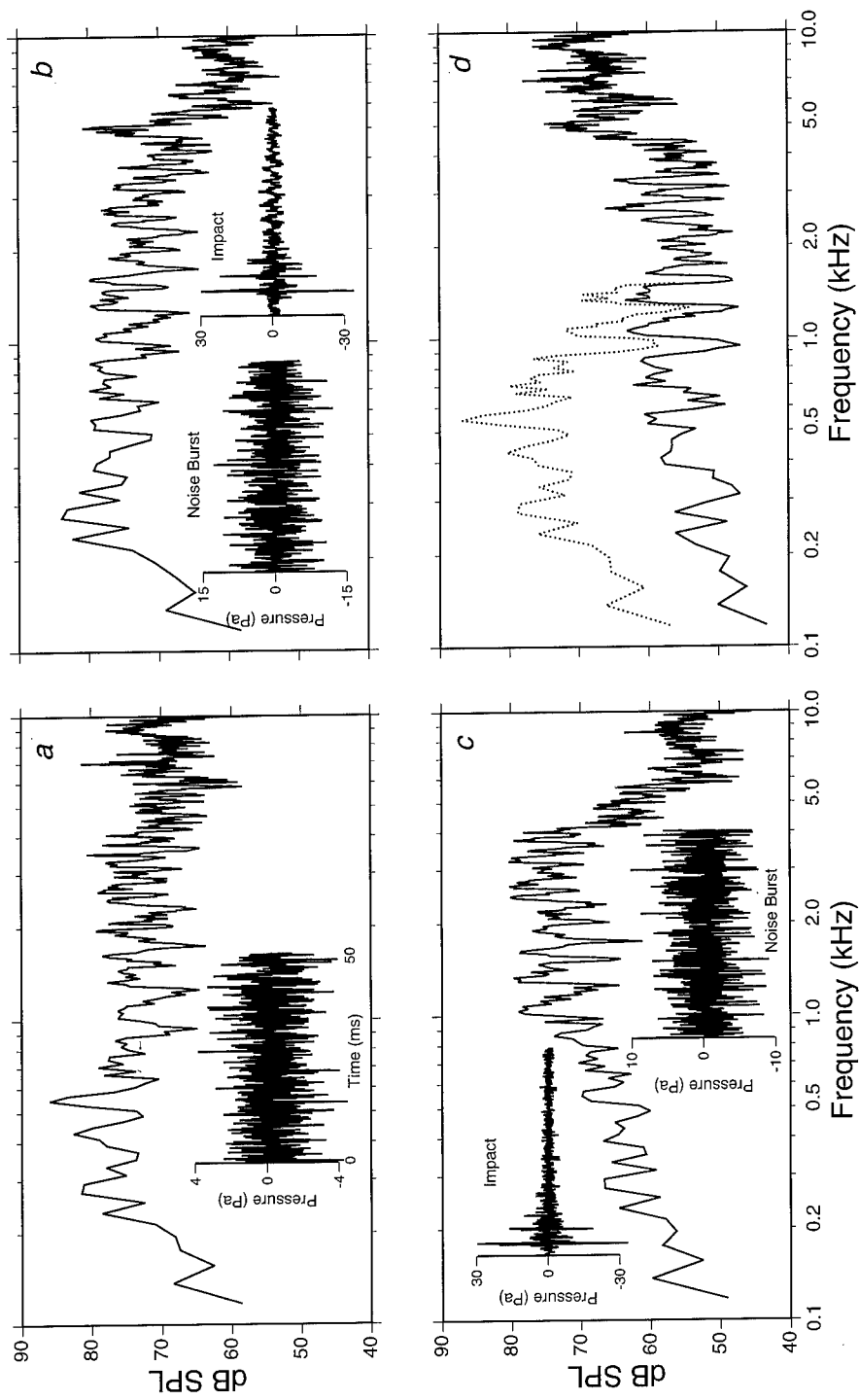


Figure 18-9 (a) Average spectrum of the broadband noise with an $L_{eq} = 100$ dB. (b) Average spectrum of the noise burst and impact shown in the inset for the $\beta(t) = 21$ and $\beta(t) = 84$ noises. (c) Average spectrum of the noise burst and impact shown in the inset for the $\beta(t) = 10$ and $\beta(t) = 45$ noises. (d) Average spectra of the Gaussian noises (solid line) that were combined with the transients in panel (b) and (dotted line to 1.5 kHz and solid line from 1.5 to 10 kHz) with the transients of panel (c) to produce the four non-Gaussian noise exposures.

ond sample of the digitized temporal waveform (i.e., over 5000 successive windows). Although $\beta(t)$ is a statistically based metric that quantifies the deviation of the amplitude distribution of a temporal signal from the Gaussian, the frequency domain kurtosis, $\beta(f)$, provides a similar index but of the temporal fluctuations of a given frequency component or band of the spectrum over time. The $\beta(f)$ was computed on selected octave bands of the digitally filtered temporal samples over a 51.2 second period, (i.e., over 1000 windows). The samples of every window were convolved with the impulse response of the octave bandwidth filter. The filter was designed as a finite impulse response (FIR) digital filter in which the coefficients of the impulse response were obtained from the DFDP3/plus filter package

(Atlanta Signal Processing, Inc.). The length of the FIR filter was 2000 and the total length (4096) of the array or basis (size of the memory locations) performing the convolution was four times as long as the length of a window because the convolution of digital samples has the summed duration of the length of the input samples and the length of the impulse response. To avoid temporal aliasing the convolution was performed on a basis having a longer length (4096) than the length of the input samples (1024) plus the length of the impulse response (2000). The first 1024 samples of the results of the convolution were preserved as the samples of the filtered output. The filtering process was performed repeatedly to obtain $\beta(f)$ over successive octave bands.

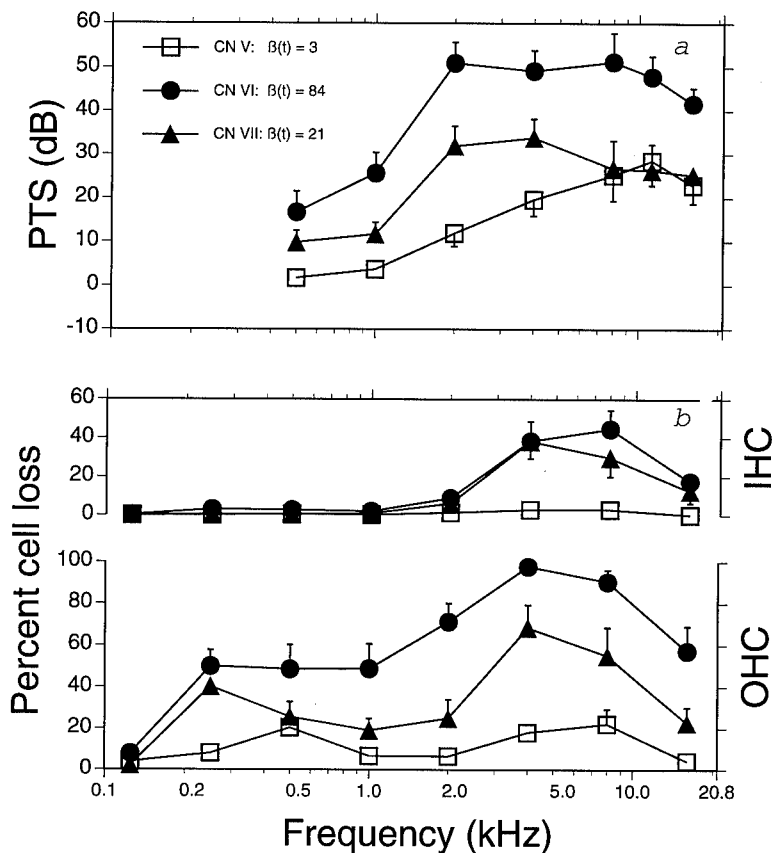


Figure 18-10 (a) Mean permanent threshold shift (PTS) audiograms and (b) inner and outer hair cell loss (IHC, OHC) for the groups of animals exposed to the 100 dB SPL broadband noises having the indicated time domain kurtosis, $\beta(t)$.

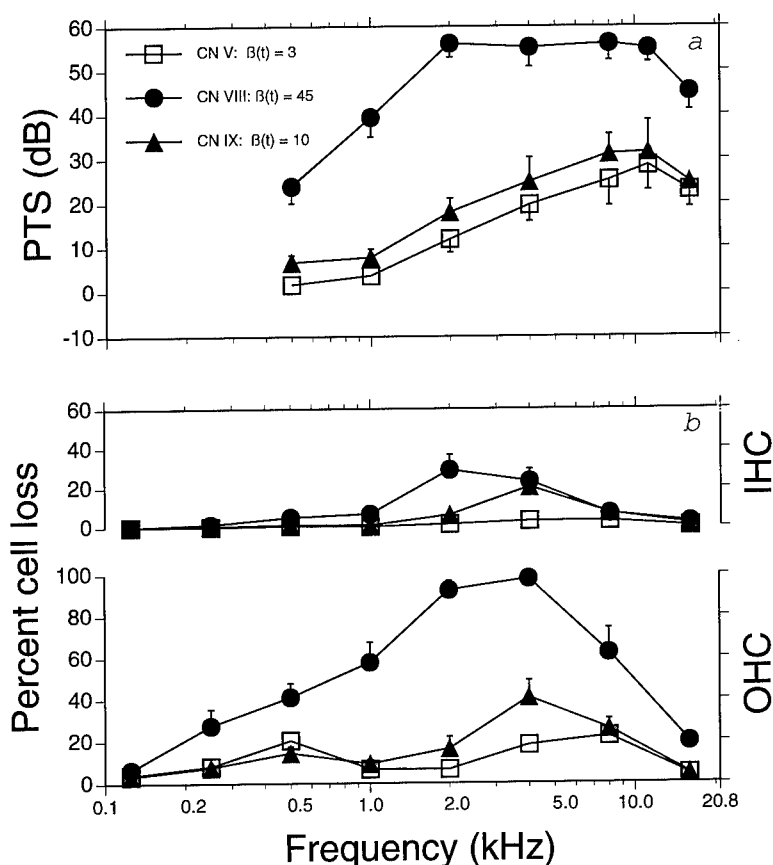


Figure 18-11 (a) Mean permanent threshold shift (PTS) audiograms and (b) inner and outer hair cell loss (IHC, OHC) for the groups of animals exposed to the 100 dB SPL broadband noises having the indicated time domain kurtosis, $\beta(t)$.

Thus, a series of exposures were created, all having the same L_{eq} and the same long-term spectrum, but differing quite dramatically in their temporal structure. The results of exposure to these five exposure paradigms are shown in Figures 18-10 and 18-11. Clear and dramatic differences were seen in the magnitude and frequency specificity of the PTS, OHC loss, and IHC loss. The frequency specificity of the loss reflects, in part, the low-frequency mechanical transmission characteristics of the chinchilla ear as well as the effects of the transient frequency changes that were designed into each of the noise exposures. There are different audiometric and histological consequences for all five of these exposure conditions, despite all exposures having the same

spectrum and L_{eq} . The audiometric and histological results for each of the five exposures shown in Figures 18-10 and 18-11 are replotted in Figures 18-12 and 18-13 as the difference in PTS or OHC loss between the indicated exposure and $\beta(t) = 3$, Gaussian exposure, along with the octave band $\beta(f)$ distribution. Several features of the $\beta(f)$ metric are clearly evident: the $\beta(f)$ "spectrum" generally follows the frequency profile of audiometric threshold shift as well as the profile of OHC loss; OHC loss is consistently shifted approximately an octave relative to the $\beta(f)$ profile; in the $\beta(t) = 10$ exposure the relatively small difference value for PTS across frequency is not reflected in the difference value for OHC loss, but the $\beta(f)$ function clearly reflects the OHC pathology.

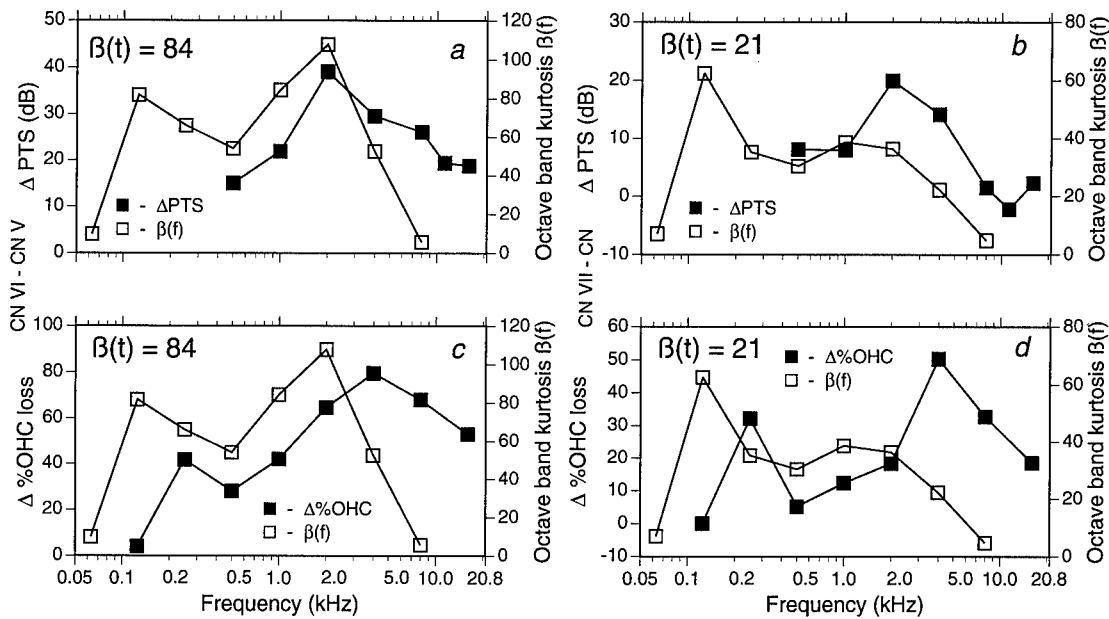


Figure 18-12 The difference in mean permanent threshold shift (Δ PTS) and percent outer hair cell loss ($\Delta\%$ OHC) between the groups exposed to the indicated $\beta(t)$ exposure and the $\beta(t) = 3$ (Gaussian) noise exposure compared with the (frequency domain kurtosis) $\beta(f)$ spectrum.

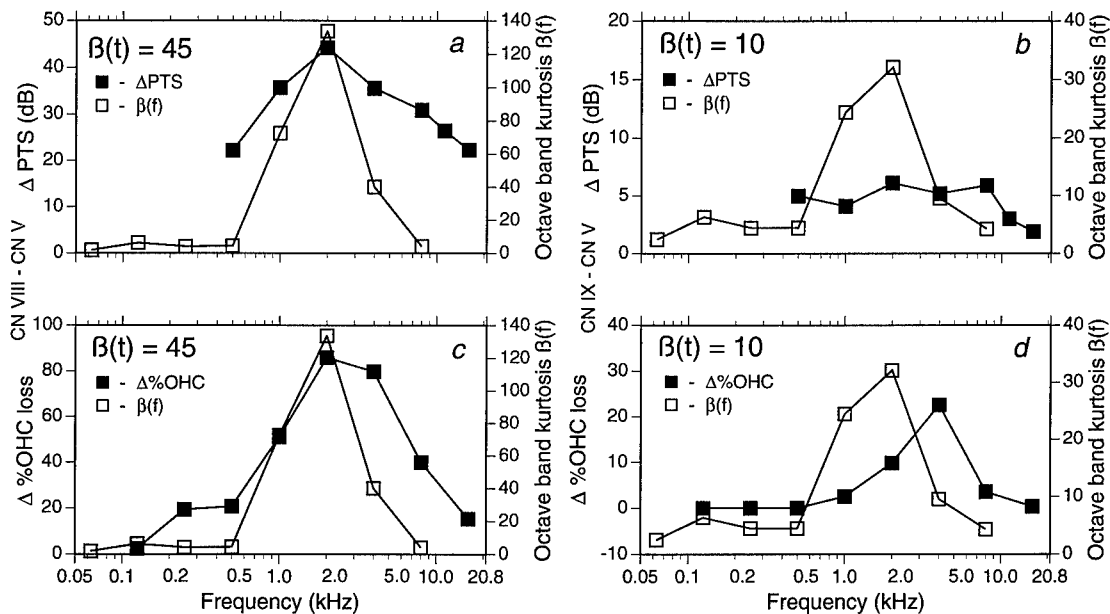


Figure 18-13 The difference in mean permanent threshold shift (Δ PTS) and percent outer hair cell loss ($\Delta\%$ OHC) between the groups exposed to the indicated $\beta(t)$ exposure and the $\beta(t) = 3$ (Gaussian) noise exposure compared with the (frequency domain kurtosis) $\beta(f)$ spectrum.

Figure 18-14 shows the relation between the $\beta(t)$ value for each of the five exposure paradigms and the group mean total loss of OHCs throughout the cochlea. Clearly seen in this figure is the systematic relation between the two variables. Both $\beta(t)$ and $\beta(f)$ correlate well with the audiometric and histological results and, while both metrics rank order the magnitude of trauma, $\beta(f)$ is also remarkably suggestive of the frequency specificity of the trauma, especially OHC loss.

Taken together these experiments show that the temporal structure, of an exposure, is an important variable in determining trauma, and that conventional measures of an exposure such as L_{eq} are not sufficient to predict the extent of the trauma. However, if an efficient algorithm for obtaining $\beta(f)$ can be developed, the kurtosis metric holds promise for predicting the magnitude and frequency profile of the resultant trauma, especially if the algorithm is built in such a way that the transfer characteristics of the external and middle ear are taken into account.

Concluding Comments

The analyses of the permanent audiometric and histological effects of the exposures described have shown that there were differences among the five experimental groups that were produced by the non-Gaussian structure of the noises. Consistent differences in PTS of more than 40 dB at some frequencies and very large differences in the sensory cell losses within some octave band lengths of the basilar membrane were found between the effects of spectrally and energy equivalent Gaussian and non-Gaussian continuous noise exposures. The fact that temporal-spectral variables are important determinants of hearing loss is not surprising because the cochlea has evolved to transduce nonstationary, stochastic signals. The signal is transformed in the cochlea into a frequency representation along the basilar membrane, and this representation is continually changing in time. Thus it seems reasonable that analytical methods for transforming a nonstationary signal, having features in common with the cochlear trans-

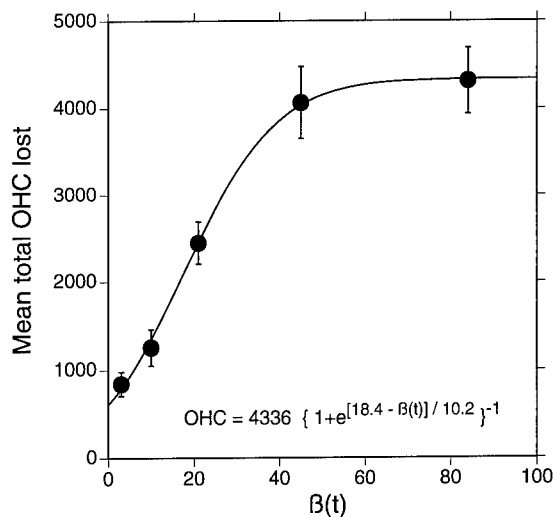


Figure 18-14 The relation between $\beta(t)$ and the group mean total number of outer hair cells (OHC), lost following exposure to the various 100 dB SPL broadband noises. The solid line is a curve fit to the data given by the equation shown.

duction process (i.e., a joint time-frequency representation), might be good prospects for use in extracting information from the signal that could be used to quantify the signal's potential for producing hearing loss. If a stochastic signal is nonstationary, a Fourier-based spectral description is no longer time independent and alternate methods for studying temporal variations in spectral properties are required. An algorithm developed by Dwyer²⁴ was initially used to obtain $\beta(f)$. The results proved difficult to relate to the audiometric and histological data because the Dwyer algorithm yields values of $\beta(f)$ on a uniformly scaled frequency dimension while the audiometric and histological data reflect processes occurring on a logarithmic scale. The most obvious alternative was to use a straightforward filtering approach to obtain $\beta(f)$. However, with our computer system, the method was extremely time consuming and required over 8 hours of computer time to obtain the $\beta(f)$ function for a given noise environment. Thus, although the method used to obtain the data presented in Figures 18-12 and 18-13 was inefficient and difficult to implement as a routine measurement, the results as

indicated were enlightening. There is a consistent relation between $\beta(t)$ and the total loss of OHCs as well as a systematic relation between $\beta(f)$ and the frequency-specific profile of OHC loss across the basilar membrane as a result of the five different exposures. However, because of the nature of the kurtosis statistic, it must be used in conjunction with an energy metric to be useful in predicting the magnitude of trauma.

Obviously an algorithm for the joint frequency-time representation of a signal that could operate on any scaling of frequency would be needed if the $\beta(f)$ metric is to be efficiently obtained. The wavelet transform (WT) is such an analytical procedure and is being developed for the efficient extraction of $\beta(f)$. WT is an analytical technique for characterizing a signal in the time-frequency plane thus combining time and frequency analysis. The WT has the additional advantage over other joint time-frequency representations such as the short-time Fourier transform in that it can operate on a nonlinear frequency scale.

Acknowledgments

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Animal Use

In conducting the research described in this study, the investigators adhered to the *Guide for the Care and Use of Laboratory Animals* prepared by the Committee on Care and Use of Laboratory Animals of the Institute of Laboratory Animal Resources, National Research Council [DHHS Publication No. (NIH) 86-23, revised 1985].

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Chapter 19

Fetal Response to Intense Sounds

Kenneth J. Gerhardt, Linda L. Pierson,
and Robert M. Abrams

The fetus develops in a dynamic environment that provides stimulation and nutrition, yet at times can be potentially hazardous. For example, drugs ingested by the mother can be passed to the fetus and may produce devastating consequences. A variety of other agents external to the fetal niche, such as sounds and vibrations, impacts the organism directly. Knowledge of fetal reactions to environmental sounds has stimulated interest in prenatal learning, and at the same time, created concern that the fetal auditory system may be adversely affected by intense noise. Exposure to maternal vocalization during prenatal life may contribute to speech perception and voice recognition by the newborn.¹ The outcome of these early experiences may prove beneficial later in life. On the other hand, noise exposures that are hazardous to the hearing of adults may be hazardous also to the hearing of the fetus.²

In this chapter we review recent developments in our understanding of the sounds that reach the fetus and the influence these sounds may have on the organism. Topics include the fetal sound environment, sound penetration to the fetal head, fetal response to exogenous sounds, exposures to hazardous noises, and temporary noise-induced hearing loss identified in fetal sheep.

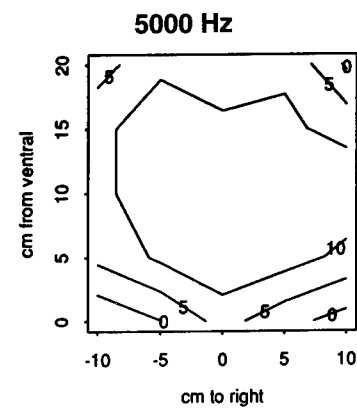
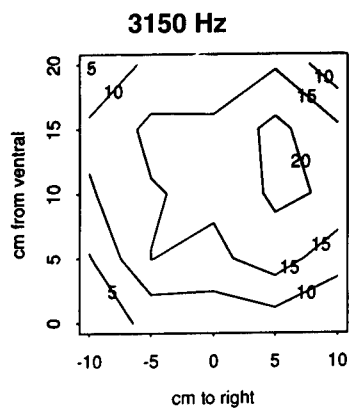
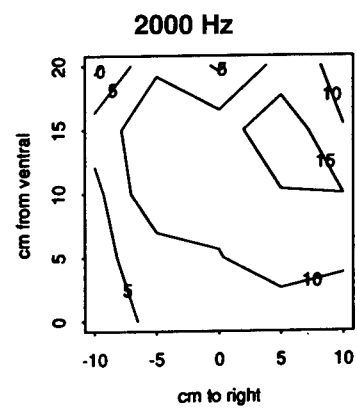
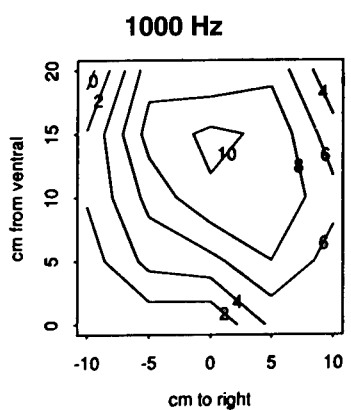
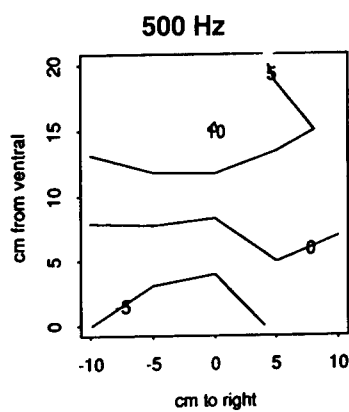
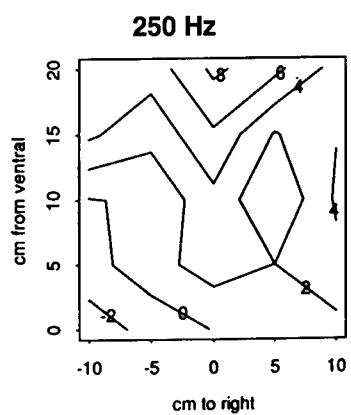
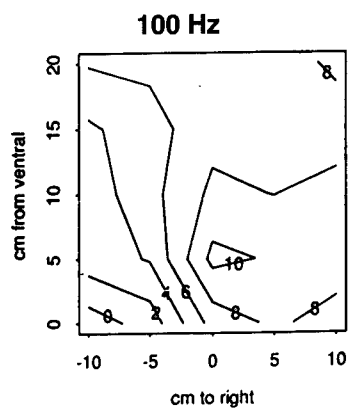
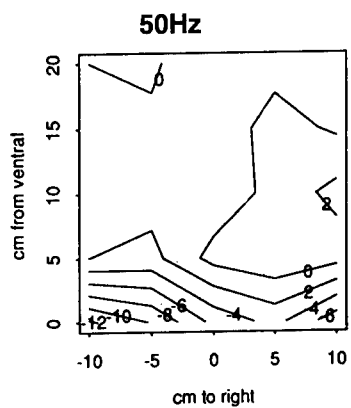
Sound Transmission into the Uterus

The fetal sound environment is composed of a variety of internally generated noises, as well as many sounds originating from the environ-

ment of its mother. The once held belief that the fetus develops in an environment devoid of exogenous stimulation has been eroded by information showing consistent fetal responses to sound. Specifications of the amplitudes and frequency distributions of sounds reaching the fetal head have implications for our understanding of fetal responses. The stimulus used to produce a fetal response is altered as it passes from an air medium through the abdominal wall and uterus and into the amniotic fluid. This topic has been reviewed recently by Gerhardt³ and Busnel et al.⁴

The acoustic characteristics of internal noises and of external sounds that penetrate the uterus have been described in the human from various recording sites including inside the cervix⁵ and inside the uterus after amniotomy.^{6,7} These intrauterine sounds were very similar to those recorded in pregnant sheep via a chronically implanted hydrophone on the fetal head inside the intact uterus.⁸⁻¹⁰ The general agreement among recent studies using both humans and sheep, in part influenced by similarities in the dimensions of the abdomen during pregnancy, supports the continued use of sheep as a model for describing sound transmissibility from exogenous sources into the uterus.

Sounds generated inside the mother and present in the uterus are associated with maternal respiratory, cardiovascular, intestinal, and laryngeal activity, and by physical movements.¹⁰⁻¹² These sounds provide a background above which maternal vocalizations and externally generated sounds emerge. In-



ternal sounds are predominately low frequency (<100 Hz) and reach 90 dB sound pressure level (SPL, re: 20 μ Pa).¹⁰ Spectral levels decrease as frequency increases, and are as low as 40 dB for higher frequencies.¹³

Exogenous low-frequency sounds less than 250 Hz penetrate the uterus with very little reduction in sound pressure (<5 dB). Some enhancement of low-frequency sound pressures has been reported in both humans¹⁴ and sheep.^{8,10} In other words, sound pressures can be greater inside the abdomen than they are outside the abdomen. Higher frequencies up to 5000 Hz are attenuated by approximately 20 dB. These general findings have been refined and extended by Peters et al.,¹⁵ who evaluated the transfer of airborne sounds across the abdominal wall of sheep as a function of frequency and intraabdominal location.

In the study by Peters et al.¹⁵ a hydrophone was positioned at each of 45 locations in the abdomen of five sheep. A loudspeaker was located 1 m away from the left flank and broadband noise was delivered to the side of the ewe at 90 dB SPL. Spectra of the noise within the abdomen were evaluated from each hydrophone location and from a microphone positioned 10 cm away from the flank. Sound pressure attenuation was calculated as the SPL difference recorded with a microphone located by the flank and with the hydrophone at each of the 45 positions. Isoattenuation contours within the abdomen are seen in Figure 19-1 for eight selected frequencies.

The sound pressure at different locations within the three-dimensional space of the ewe was highly variable. Low-frequency bands of noise revealed strong enhancement of sound pressure by up to 12 dB in the ventral part of the abdomen. For midfrequencies (250–2000 Hz), attenuation reached as high as 20 dB. Attenuation for higher frequencies (>3150 Hz) were somewhat less than for midfrequencies. Figure 19-2 includes plots of the attenua-

tion by frequency for hydrophone locations from the left flank through the center to the right flank. This general pattern of sound attenuation has been observed in both humans¹⁴ and sheep.^{8,10}

Over the frequency range from 125 to 2000 Hz, the abdomen can be characterized as a low-pass filter with a rejection rate of approximately 6 dB/octave.¹⁰ Thus, external stimuli are shaped by the tissues and fluids of pregnancy before reaching the fetal head. This shaping should have considerable effect on the subsequent signal to which the fetus is exposed.

Response of Fetus to Exogenous Sounds

The human developmental and obstetrical literature includes numerous studies of the response patterns of acoustically stimulated near-term fetuses. Fetal heart rate responses, body and limb movements, and fetal eye blinks are common indices of fetal responsiveness to both high- and low-intensity airborne sounds.^{4,16,17} Various transducers have been used to elicit fetal responses including loudspeakers, earphones, door bells, etc. placed near the abdomen. Specifications of stimulus levels and spectral content are lacking in most reports. However, studies that provide clear stimulus parameters, for example Lecanuet et al.,¹⁷ revealed that intense airborne stimulation (above 105 dB SPL) produced short-latency fetal heart rate accelerations accompanied by motor responses. Less intense stimulation (below 100 dB) still produced heart rate accelerations, but fewer motor responses. When the amplitude or the frequency of the stimulus increased, motor and cardiac accelerations increased during either quiet or active sleep.¹⁸ The magnitude of the response is influenced by the behavioral state of the fetus during the time of stimulation. Schmidt et al.¹⁹ found that stimulation produced stronger responses

Figure 19-1 Cross-section isoattenuation contours within sheep abdomen. Airborne broadband noise was delivered to left flank and recorded intraabdominally. Attenuation contours of eight selected frequencies are displayed. Negative values indicate sound enhancement. Reprinted with permission from Peters et al.¹⁵

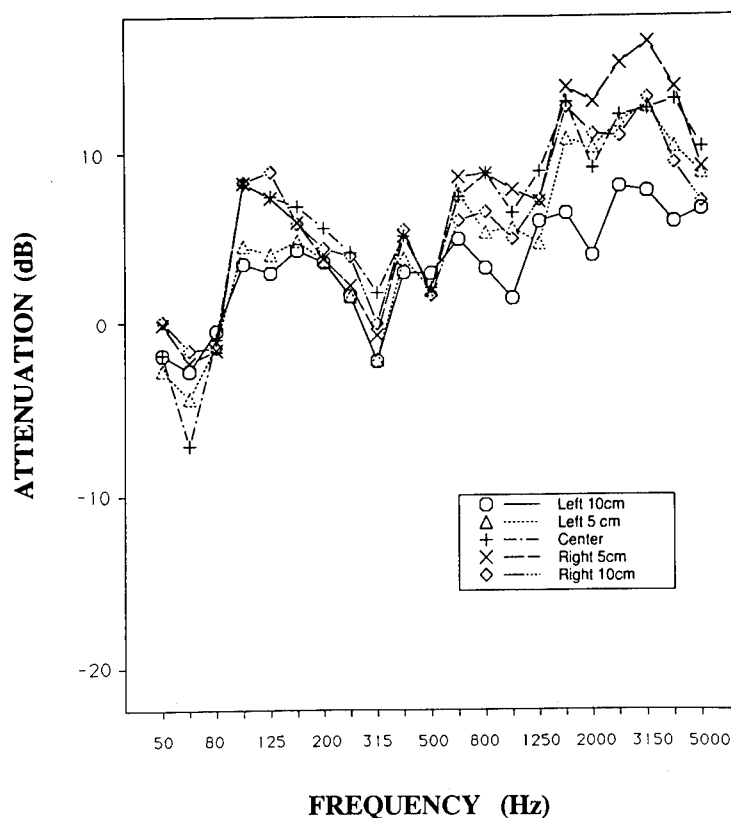


Figure 19-2 Mean sound attenuation as function of frequency for hydrophone locations in left-right dimension (cross section). Data were collapsed over heights ($n = 5$) and animals ($n = 5$). Reprinted with permission from Peters et al.¹⁵

during quiet and active wakefulness than during the different stages of sleep.

Obstetricians interested in the well-being of the fetus have searched for reliable, noninvasive procedures to assure fetal viability. Important signs of fetal distress are reduced body and limb movements and loss of heart rate accelerations. Unfortunately, these characteristics are present also in healthy, quiet fetuses. Thus, the physician may attempt to arouse the fetus by direct manipulation of the maternal abdomen or, more recently, by stimulating the fetus with a vibroacoustic signal. In healthy fetuses, vibroacoustic stimulation will usually result in fetal movements and cardioaccelerations.^{20,21} Fetal acoustic stimulation testing using the electrolarynx, a device designed to assist alaryngeal talkers, has been introduced in clinical practices for antenatal and intrapartum surveillance.²² The electro-

larynx, a convenient sound source, is now a commonly used stimulator for obstetrical examinations.

One disadvantage of the electrolarynx is the high stimulus level produced within the uterus.²³ The electrolarynx produces SPLs up to 129 dB in the human uterus near the fetal ear^{6,24} and up to 135 dB in the sheep uterus.²³ Occasionally, an exaggerated response from the human fetus has been observed, suggestive of discomfort and perhaps even pain.²¹ Stimulation during labor has resulted in cases with prolonged fetal bradycardia,²⁵ which in one case necessitated a caesarean section.²⁶

Figure 19-3 shows the SPL of the electrolarynx recorded in the uterus of a pregnant ewe as a function of the distance between the stimulator and the recording hydrophone.²³ At the center location (electrolarynx directly over the hydrophone), SPL averaged 134.9

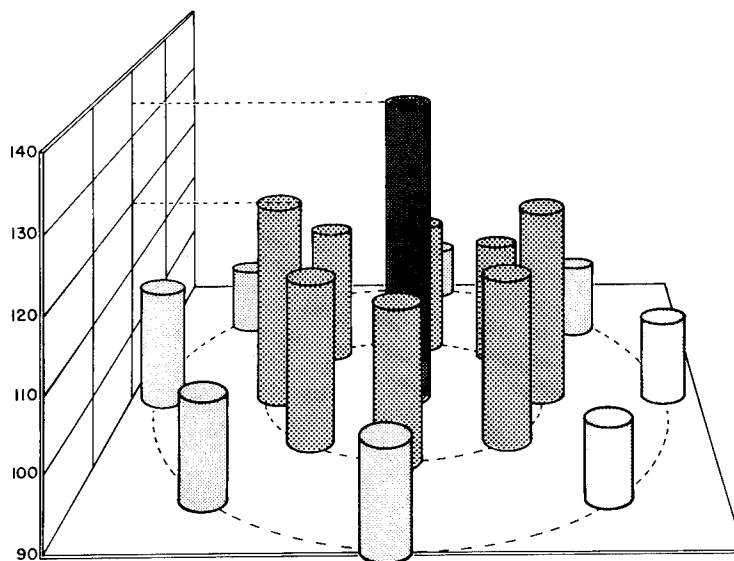


Figure 19-3 Graphic representation of the average sound pressure (in decibels) generated by the electronic artificial larynx at various positions on the abdomen. Reprinted with permission from Gerhardt et al.²³

dB. At 3 and 6 in, the average SPLs were 114.3 and 101.4 dB, respectively. Also, signal levels tended to be lower the deeper the hydrophone was within the amnion. The level of stimulation in the uterus depends on many factors including the distance between the vibrator and the recording transducer, the force of application of the electrolarynx on the abdomen, and the frequency content of the device.

Whether or not hearing loss can be induced by a brief, one time exposure to these signals has been addressed in a few recent articles. The hearing of young children who were exposed to vibroacoustic stimulation testing during fetal life were tested using auditory brain stem response (ABR)²⁷ and pure-tone audiometry.²⁸ Arulkumaran et al.²⁸ screened 465 4-year-old children at 25 dB and found two cases of hearing loss. Causes for the hearing losses were unknown. Ohel et al.²⁷ tested 20, 1–2-day-old infants and found no ABR latency differences when compared to a control group of age-matched infants. Both reports concluded that in utero exposure to the electrolarynx had no effect on the hearing of these children. The likelihood of brief, intense exposures producing hearing loss in the fetus is not

great. However, pure-tone screening at 25 dB or using screening ABR would not provide adequate sensitivity to rule out subtle changes in hearing.

Local Cerebral Glucose Utilization

Another method for evaluating effects of sound stimulation on in utero fetuses involves assessing local cerebral glucose utilization in animals. Sheep^{29,30} and guinea pigs³¹ have been used for these studies. During stimulation, energy metabolism increased in brain stem auditory structures of guinea pigs and in all auditory structures including the auditory cortex of sheep. In fetal sheep with ablated cochleae, there was a widespread decline in glucose utilization even during sound stimulations.³⁰ Stimulation of in utero guinea pigs to intense pure tones delivered through a loudspeaker produced evidence of tonotopic organization in the cochlear nucleus and in the inferior colliculus. There was a marked increase in metabolic activity of isofrequency populations of cells in the fetus in response to tonal stimulation delivered to the pregnant

guinea pig. This topic has been reviewed by Granier-Deferre and Abrams.³²

Sound Isolation of Fetus

Human fetal auditory responsiveness begins about the 24th week of gestation.^{33,34} During the next 15 weeks exogenous sounds may have an effect on fetal behavior and central nervous system development. The positive benefits of sound, for example speech perception and voice recognition in the newborn, may result from direct stimulation of prosodic features of the maternal voice heard by the fetus prenatally.¹ On the other hand, intense sounds such as those found during vibroacoustic stimulating testing evoke atypical changes in fetal behavioral state and fetal movements that persisted long after stimulation.²¹

Our understanding of the possible benefits of sounds during prenatal life, as well as the possible adverse effects of intense sound exposures, is incomplete. We have a fair idea about how much sound pressure is present at the fetal head and now have information about how much sound actually reaches the fetal inner ear.³⁵

Inferences regarding sound transmission to the inner ear can be made from cochlear microphonic (CM) input-output functions to stimuli with different frequency content. The CM, an alternating current generated at the hair cells, mimics the input signal in frequency and amplitude over a fairly wide range. CMs recorded from the round window are sensitive indices of transmission characteristics of the middle ear. Thus, changes in the condition of the middle ear influence the amplitude of the CM. Comparisons of CM recorded from in utero fetuses to sound field stimulation and CM recorded from young lambs after delivery in the same sound field provide estimates of fetal sound isolation.

During sterile surgery in sheep, Gerhardt et al.³⁵ exteriorized the fetus and instrumented it with round window recording electrodes. Electrode leads were passed through the maternal uterus, abdomen, and skin, and secured in a pouch sutured to the flank of the

ewe. At least three days after surgery, CM input-output functions were recorded to 1/3-octave bands of noise (centered at octave intervals from 0.125 through 2.0 kHz) delivered through a loudspeaker located 1.8 m from the ewe.

After recording from the fetus, the animal was delivered by cesarean section. At least 24 hours passed, during which time fluid in the middle ear was presumably drained. CM input-output functions produced by the same stimuli were repeated in the same sound field. The difference between the SPL necessary to generate a fixed CM amplitude from the in utero fetus and the SPL necessary to produce the same amplitude from the animal after delivery served as an index of fetal sound isolation.

Input-output functions for nine fetuses and six newborn lambs were analyzed. Within each preparation, a comparison of CM using these input-output functions from each condition (fetus and newborn) was completed. Figure 19-4 illustrates the method for calculating fetal sound isolation. The CM functions

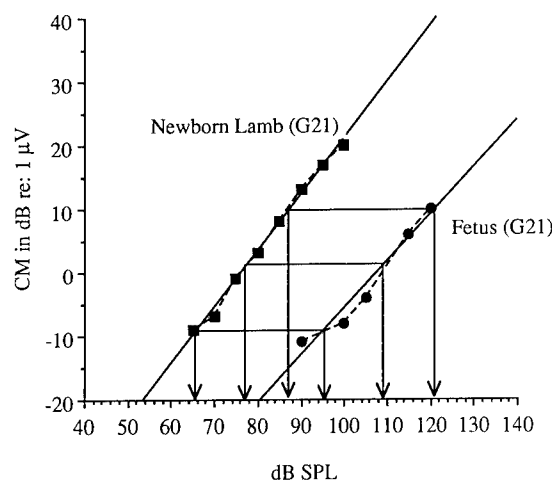


Figure 19-4 Fetal and newborn CM input-output functions fitted with linear regression lines. The distance between the regression lines (expressed in decibels along the x-axis) was the average of three measurements (31.3 dB) and represented fetal sound isolation from animal G21 at 1.0 kHz. Reprinted with permission from Gerhardt et al.³⁵

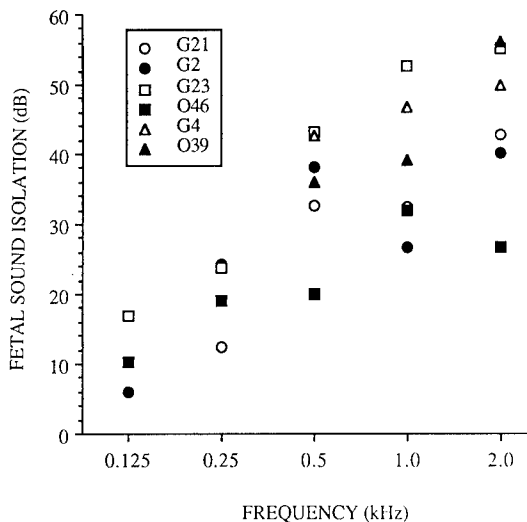


Figure 19-5 Fetal sound isolation. The average differences between the SPLs in decibels necessary to produce equal cochlear microphonic (CM) function for each frequency recorded from the fetus and the newborn (ex utero CM minus in utero CM). Reprinted with permission from Gerhardt et al.³⁵

from both conditions for a particular stimulus frequency were plotted together and fitted with linear regression lines. The authors assumed that the CM functions were linear from a point above the noise floor to below the point of rollover. For the most part, this assumption was accurate. However, not all CM regression lines were parallel. Therefore, the difference between the regression lines was determined by averaging three points along the nearly parallel lines. The average difference in decibels (x-axis) served as the amount of fetal sound isolation for that frequency. For example, as illustrated in Figure 19-4, sound isolation was 31.3 dB at 1.0 kHz. The process was repeated for each preparation and the results appear in Figure 19.5.

The amount of fetal sound isolation was dependent upon stimulus frequency. For 0.125 kHz, sound isolation ranged from 6 to 17 dB, whereas for 2.0 kHz, fetal sound isolation ranged from 27 to 56 dB. Thus, the fetus appears well isolated from intense sounds at and above 500 Hz. However, for 0.125 kHz, fetal sound isolation averaged 11 dB.

At least two factors influence the stimuli that evoke a response from the fetus. First, the amount of sound pressure attenuation by frequency provided by the tissues and fluids surrounding the fetal head determines the spectral shaping of the signal. The energy carried by these pressure variations becomes an important factor in stimulating fetal hearing. In water, as in air, an acoustic field can be described both as a pressure variation within the medium and as a back and forth oscillation of the component particles of the medium. The two quantities, pressure and particle velocity, are related and are dependent on the acoustic impedance of the medium, that is, its density and elasticity. The acoustic impedance of water is much higher than air and for a given pressure disturbance, the particle velocity is much less, by a factor of approximately 3600 or about 35 dB.³⁶ Thus, one would assume that the stimulus level required to produce a physiological response from the fetus would be 35 dB greater than the stimulus level in air necessary to produce the same response from the newborn.³⁵ This was not the case for low-frequency sounds (<0.5 kHz) with wavelengths larger than the dimension of the ewe's abdomen.

A second important factor is the transmission of sound pressure from the fluid at the fetal head into the fetal inner ear. Transmission is governed by the route that pressure variations take to reach the inner ear. As yet, we know little about this from an experimental perspective. However, Querleu et al.³⁷ speculated that sound pressures in the fetal environment may pass easily into the inner ear through the external canal and middle ear because the impedance of the fluids found in the ear canal and middle ear are the same as the impedance of fluids found in the inner ear. Thus, middle ear amplification provided by the ossicular chain would not be necessary for efficient sound delivery to the cochlea according to Querleu's reasoning. A contrasting hypothesis consists of in utero hearing via bone conduction or through a combination of routes. Currently, experimental data describing the route of stimulus transmission into the fetal inner ear are not available.

Noise-Induced Shifts in Fetal Sheep ABR

The possibility of noise-induced, prenatal hearing loss, with or without regard for a critically susceptible period for noise damage, is of practical as well as scientific interest. There are only a few studies that have addressed this issue in either pregnant animal models^{38–40} or in children whose mothers were exposed to industrial noise during their pregnancy.^{2,41,42}

Cook et al.³⁸ exposed a group of guinea pigs during their last trimester of pregnancy to tape-recorded loom noise at 115 dBA. After delivery, ABR latencies recorded from the young guinea pigs were compared to similar measures obtained from control animals. Prolongation of wave IV latencies in the noise-exposed animals was found and attributed to overstimulation of the auditory system during development. The susceptibility of the immature auditory system to overstimulation seems to be greatest during or immediately after the period of rapid auditory maturity.^{43–46}

Using an animal model that has acoustic transmission characteristics into the uterus similar to that in humans, Dunn et al.³⁹ repeatedly exposed pregnant sheep to 130 dB SPL broadband noise for 4 hours a day. After delivery, measurements of ABR thresholds from 30- to 40-day-old lambs were not significantly different from control animal thresholds. Serial celloidin sections revealed twice as much cellular change in the noise-exposed group, but preparation artifact could not be ruled out.

Support for the possibility that noise exposures during fetal life result in changes in hearing sensitivity was provided in demographic studies of children whose mothers were exposed to noise during pregnancy.^{2,41,42} Both research teams found an increased risk of hearing loss in children whose mothers were occupationally exposed to hazardous noise levels. There was a significant increase in threshold to 4000 Hz pure tones when the exposure included a strong component of low-frequency noise. Although both studies lacked adequate control groups and found relatively small degrees of hearing loss,⁴⁷ the

possibility of fetal noise-induced hearing loss produced by intense noise exposures was suggested.

An experiment reported by Griffiths et al.⁴⁰ evaluated the effect of an intense noise exposure on the ABRs of in utero fetal sheep. Sheep are excellent animal models for this type of study because they tolerate chronic surgery well, develop hearing prenatally, and have auditory sensitivity similar to that of humans.⁴⁸ Nine pregnant ewes carrying fetuses with gestational ages between 124 and 129 days (average gestational period is 145 days) were prepared for sterile surgery following standard protocols. The animal was anesthetized, the abdomen and uterus were incised, and the fetal head was exteriorized. The fetal skull was exposed, stainless steel screw electrodes were secured at the vertex and at both mastoids, and a bone oscillator was fixed on the occipital bone. The scalp incision was closed and the fetus was returned to the uterus. The uterus and abdomen of the ewe were closed and the electrode leads and bone oscillator wire were passed through the maternal flank and stored in a pouch sutured to the side of the ewe.

After recovery from surgery, the ewe was placed in a cart and wheeled into a sound-

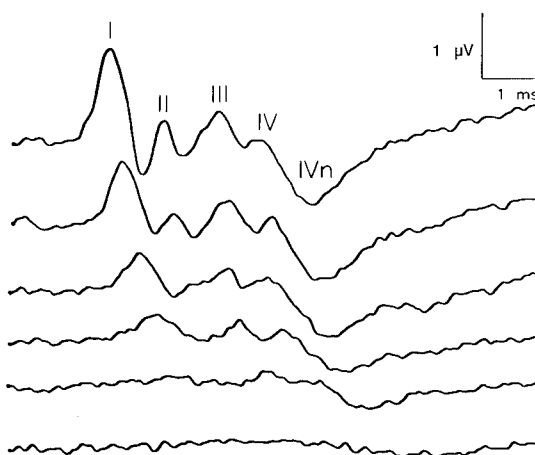


Figure 19-6 Examples of click-evoked ABR waveforms. Stimulus levels used were (from top to bottom) 41, 31, 21, 11, 1, and -9 dB normal Hearing Levels (nHL). Reprinted with permission from Griffiths et al.⁴⁰

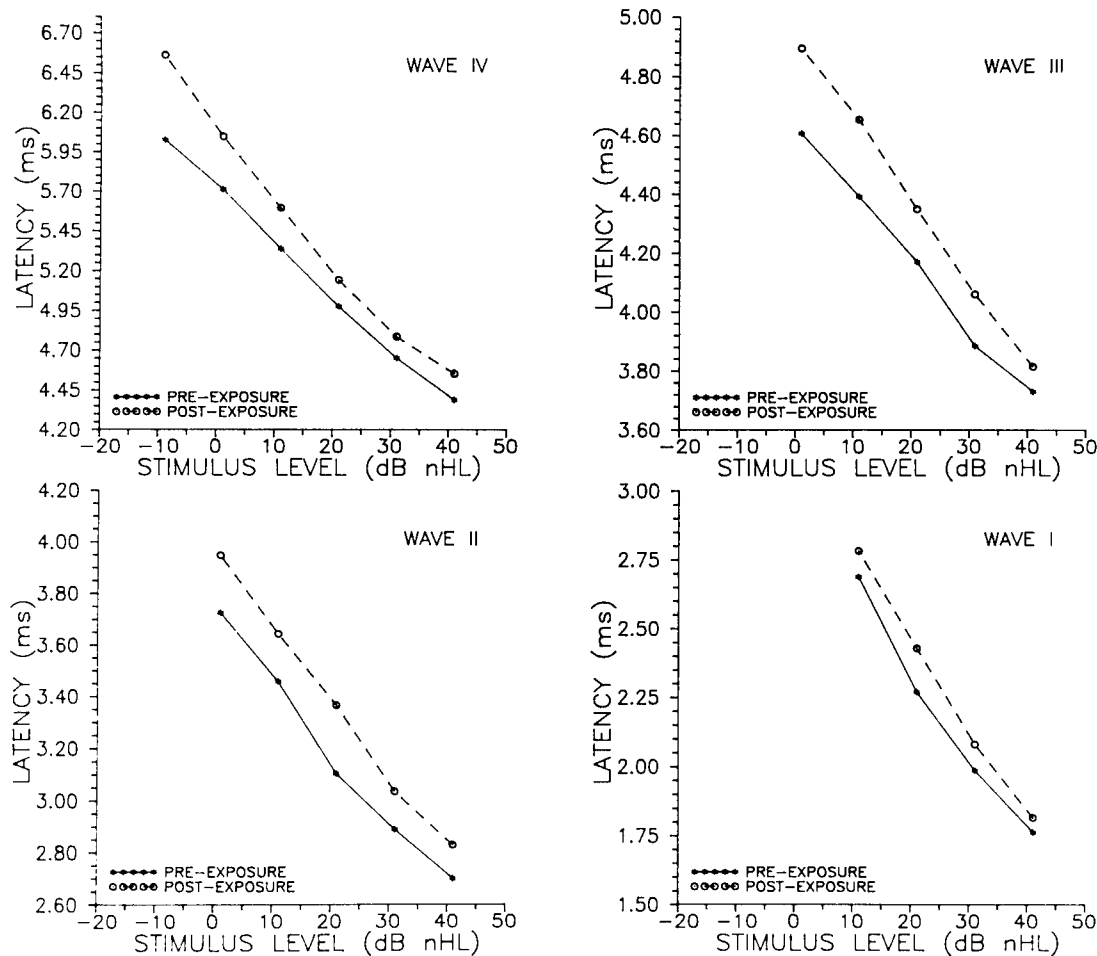


Figure 19-7 Click-evoked ABR latency-intensity functions for waves I-IV in the pre- and postexposure test conditions. Reprinted with permission from Griffiths et al.⁴⁰

treated booth. The electrode leads and bone oscillator wire were connected to an evoked potential unit and ABRs were recorded to tone bursts (0.5, 1, 2, and 4 kHz) and clicks delivered through the bone oscillator. Stimulus levels were referenced to normal hearing adult subjects using the bone oscillator placed on the mastoid. Latency-intensity functions and thresholds were recorded before, immediately after a noise exposure, and 24-96 hours later. The noise exposure (120 dB SPL broadband for 16 hours) was delivered to the ewe through loudspeakers in the sound-treated booth.

Examples of the click-evoked ABRs for a range of stimulus levels are found in Figure

19-6. The preexposure ABR waveforms consisted of four vertex-positive waves, labeled I-IV. Latencies lengthened as stimulus intensities decreased and wave IV was the only identifiable wave in the response near threshold. The waveform morphologies evoked by the tone bursts, particularly wave IV, were similar to the morphology evoked by the clicks.

Mean pre- and postexposure latencies evoked by clicks are plotted against stimulus level for each of waves I-IV in Figure 19-7. Pre- to postlatency shifts can be observed as a vertical difference between the preexposure (solid lines) and postexposure (dashed lines) latency-intensity functions. The latency differences in

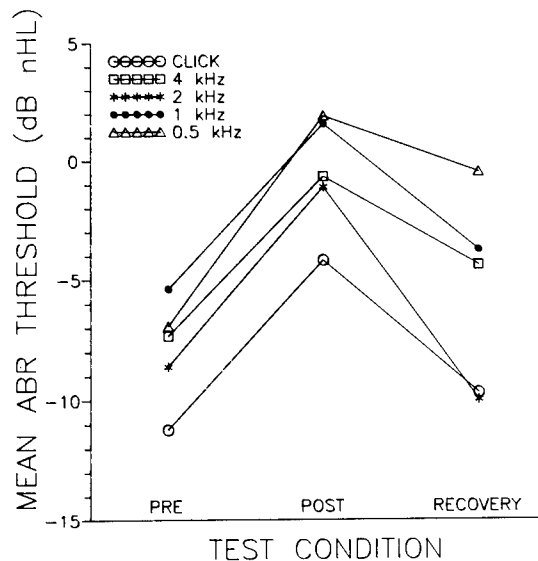


Figure 19-8 Mean ABR thresholds to clicks and 4, 2, 1, and 0.5 kHz tone bursts recorded during pre-exposure, postexposure, and recovery test conditions. Reprinted with permission from Griffiths et al.⁴⁰

these functions were significant for wave IV at all stimulus levels. Latency–intensity functions for waves III and II differed statistically at the higher stimulus levels. Latency–intensity functions for wave IV were different for the 4 kHz tone burst, but not for the lower frequencies. Following the recovery period, absolute wave latencies to all stimuli decreased to post-exposure values.

Mean ABR thresholds to all stimuli are plotted in Figure 19-8 by condition (preexposure, postexposure, and recovery). The average pre- to postexposure threshold shift was 8 dB and the average shift from postexposure to recovery was 5 dB. ABR threshold shifts noted in Figure 19-8 were statistically significant for all stimulus types except 1 kHz.

Shifts in thresholds and latencies of the ABR recorded from in utero fetuses can be produced by noise exposures delivered to the mother. The documentation of recovery after noise exposure is suggestive of temporary hearing loss. Whether or not the shifts in auditory function are accompanied by histopathologic changes in the cochlea is currently under evaluation.

Temporary ABR effects in fetal sheep whose mothers were exposed to noise has been documented by Griffiths et al.⁴⁰ Clearly, the noise exposure was of a magnitude not experienced in normal working conditions and the effects were small (8 dB). Additional information is needed regarding possible permanent effects on in utero fetuses. Moreover, evaluation of ABR threshold and latency shifts caused by overstimulation to mature sheep would provide a reference point for assessing concerns about similar exposures in the fetus. Until more information is available concerning in utero hearing loss caused by overstimulation, generalizing the results obtained from animal models to pregnant women is not appropriate.

Conclusions

Sound attenuation of exogenous signals into the uterus of sheep and of humans is small (<5 dB) for frequencies below 250 Hz and increases up to 20 dB for frequencies from 500 to 5000 Hz. Sound pressures reach the fetal cochlea and result in behavioral responses such as eye blinks, body and limb movements, and nonbehavioral responses including alterations in heart rate and increased levels of glucose utilization in the brain. The presence of fluid in the outer and middle ear spaces of fetal sheep account for further reductions in the sounds that reach the cochlea. The sheep fetus is isolated from exogenous sound with frequencies between 500 and 2000 Hz by about 40 dB, yet for lower frequencies (125 and 250 Hz) the isolation is only 10 dB. Thus, there is only a small reduction in low-frequency sound energy that reaches the fetal inner ear.

The ABRs from in utero fetuses of ewes exposed to 120 dB SPL broadband noise for 16 hours were recorded before and after the noise exposure. Presumably, this exposure results in considerable low-frequency energy in the cochlea. Statistically significant ABR threshold elevations and latency increases were identified immediately after this intense and prolonged noise exposure. Recovery of ABR thresholds and latencies occurred be-

tween 24 and 96 hours. Further information is needed regarding noise-induced threshold shifts as a function of stimulus characteristics in both fetuses and young sheep. Histologic evaluation of cochlear tissue would be of considerable value.

It must be emphasized that the results discussed above, particularly those obtained from fetal sheep, are based on limited data and must be interpreted with caution. Generalizations to pregnant women are not warranted at this time.

Acknowledgments

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Section III

**Auditory Performance Changes
With Noise-Induced Hearing Loss**

Chapter 20

Spectro-Temporal Processing in Cochlear Hearing-Impaired Listeners

Joseph W. Hall III, John H. Grose, and Lee Mendoza

Listeners with noise-induced hearing loss and other forms of sensorineural hearing loss often show marked difficulty detecting and recognizing complex signals in noisy backgrounds, and in tracking one auditory source in the presence of other competing sources. This impairment can significantly reduce the listener's ability to communicate. A large part of this difficulty can be attributed to the fact that the hearing loss renders part of the acoustic message inaudible. As suggested by Plomp,¹ another contribution to hearing disability may be that the portion of the acoustic message that is audible suffers various forms of degradation (e.g., reduced frequency or temporal resolution).

A contemporary challenge in hearing science is to determine the relationship between what is known about impairments in basic auditory abilities such as frequency and temporal resolution, on the one hand, and real-world impairments/disabilities, such as understanding speech in a noisy background, on the other. An approach that may help in this challenge is the study of psychoacoustical phenomena² that depend upon across-frequency analyses of spectro-temporal information, particularly those that may be analogous to the segregation of apparent auditory sources in multisource backgrounds. Both forms of auditory impairment (inaudible or degraded acoustical information) have potential repercussions for the way in which spectrotemporal information is processed in the synthesis of wideband signals, and in the segregation of multiple auditory sources.

There has been growing interest recently in normal auditory processes that involve the across-frequency analysis of temporal envelope information. Several new paradigms have emerged to study such processes, and it is assumed that the paradigms reveal information about peripheral and central mechanisms that ultimately enable the auditory system to assign different spectral components to particular environmental sources. Because this ability appears to be compromised in many cochlear-impaired listeners, it may prove enlightening to bring these paradigms to bear on the problem of understanding cochlear hearing loss. This chapter will discuss some possible effects of sensorineural hearing loss on auditory processes that involve across-frequency processing of spectro-temporal information.

Comodulation Masking Release

The results of many experiments are consistent with the notion that the ear is highly frequency selective.²⁻⁷ For example, when a pure tone is masked by a broadband noise, the masking effectiveness of noise components falls off steeply as the frequencies of the noise components diverge from the signal frequency. The relatively narrow region of frequencies contributing to the masking of the signal is usually called the "critical band" or passband of the "auditory filter." Interestingly, when the masking noise is not random, but has a modulation pattern that is coherent across frequency, results cannot be explained

by appealing only to the auditory filter model. When modulation is coherent across frequency, the noise energy outside the auditory filter centered on the signal frequency can actually lead to an improvement in signal detection. The improvement in detection resulting from the presence of comodulated flanking energy has been called comodulation masking release, or CMR.⁸⁻¹⁶ A model of CMR proposed by Buus⁹ hypothesizes that masking release is based primarily or exclusively upon the energy occurring in the masker dip regions of the noise. In this model, the comodulated energy serves to identify the best times to "listen" for the signal: the intervals when the flanking energy is low in amplitude. This detection strategy would improve the signal-to-noise ratio at the signal frequency. Results from several studies are in agreement with this model.^{13,16,17} Hall et al.⁸ suggested that CMR may reflect a basic auditory process that facilitates the detection and recognition of signals in modulated noise.

The available data indicate that CMR is often reduced in listeners having cochlear hearing impairment.¹⁸⁻²⁰ One issue that has received attention is whether the reductions in CMR in hearing-impaired listeners may be associated with poor frequency selectivity. Poor frequency selectivity would reduce the independence of the outputs of auditory filters centered at different frequencies, and therefore reduce the magnitude of across-frequency difference cues. Furthermore, CMR increases with the number of independent auditory filters contributing information.¹⁴ Presumably, listeners having poor frequency selectivity would have a smaller number of quasi-independent auditory filters contributing information. In support of a relationship between reduced frequency selectivity and poor CMR, Hall et al.¹⁸ and Hall and Grose¹⁹ reported modest, but significant, correlations between CMR and frequency selectivity, even when the threshold in quiet was statistically controlled. Moore et al.²⁰ reported further evidence that may support a relationship between reduced frequency selectivity and reduced CMR. Detection in modulated noise was determined as a function of increasing

noise bandwidth. In a normal-hearing listener, appreciable masking release in such noise does not occur until the noise bandwidth has exceeded the auditory filter bandwidth.¹⁴ For listeners with very poor frequency selectivity, masking release would not be expected to begin at the relatively narrow bandwidths associated with masking release in normal-hearing listeners. This, in fact, was one result obtained by Moore et al.²⁰; masking release for cochlear-impaired listeners having poor frequency selectivity did not occur typically until the masking bandwidth was relatively broad, and, even then, masking release was smaller than normal.

Another factor that could contribute to reduced CMR in cochlear-impaired listeners is poor temporal resolution. If CMR does depend upon an analysis of information in the dip portions of the temporal envelope,⁹ reduced temporal resolution could potentially limit such an analysis. Even though this hypothesis appears reasonable, actual evidence of a strong relation between reduced temporal resolution and reduced CMR is sparse. Hall and Grose¹⁹ measured temporal gap detection and CMR for the same listeners, and found only a slight correlation between poor gap detection and poor CMR. It is possible that gap detection does not reflect aspects of temporal resolution that are pertinent to the CMR task, or that in order for CMR to suffer, temporal resolution must be extremely poor (poorer than that present in the listeners sampled).

There are other factors that may lead to poor CMR in cochlear-impaired listeners. For example, high thresholds in quiet may also put a ceiling on the amount of masking release that can be obtained, due to the masked threshold being limited by the threshold in quiet.²⁰ Poor CMRs in cochlear hearing-impaired listeners may also be related to the fact that in normal-hearing listeners, CMR is smaller at low sensation levels (SLs) than at higher SLs.^{11,21} Because of loudness recruitment, cochlear-impaired listeners are generally stimulated at lower SLs than normal-hearing listeners. Therefore, it is possible that CMR is limited in cochlear-impaired listeners because of low SL. Unfortunately, matching SLs between normal-

Table 20-1 Air Conduction Thresholds for Four Subjects Having Noise-Induced Hearing Loss

	Right Ear								Left Ear							
	0.25	0.5	1.0	2.0	3.0	4.0	6.0	8.0	0.25	0.5	1.0	2.0	3.0	4.0	6.0	8.0
CC	30	35	35	35	80	85	70	40	35	25	40	45	60	65	55	45
GR	10	20	15	20	30	45	70	75	10	5	15	10	30	50	95	90
DL	25	15	25	60	50	55	80	95	20	20	20	60	60	65	100	95
WP	10	5	30	65	65	75	60	60	20	10	10	50	50	65	60	50

Bone conduction thresholds were within 10 dB of air conduction thresholds.

hearing listeners and hearing-impaired listeners has serious drawbacks: notably, loudness will generally be greater in an impaired ear, and the physiological responses of the normal and impaired ears are unlikely to be similar at equivalent SLs. It is therefore uncertain how much of a role reduced SL plays in the reduction of CMR in hearing-impaired ears. When ears are matched in terms of relatively high sound pressure level (SPL; where loud is probably similar for a normal ear and an ear with mild/moderate sensorineural loss), CMR is still generally reduced in impaired ears. Reasons other than low SL therefore probably contribute to reduced CMRs in cochlear-impaired ears.

Particular effects of noise-induced hearing loss on CMR are not presently clear because most studies investigating CMR in hearing-impaired listeners have not directly examined the effect of hearing loss etiology. We therefore present here preliminary data on a small set of listeners with the specific etiology of noise-induced hearing loss. Because noise-induced hearing loss tends to initiate in the higher frequencies, with regions of normal acuity in the lower frequencies, one general strategy was to obtain results for a frequency region where hearing was relatively normal, and results for a frequency region where hearing was relatively poor.

In the region of relatively good hearing, we used two masking stimuli: a 20 Hz wide noise band centered on 1000 Hz (on-signal band, OSB) and the OSB plus two 20 Hz wide comodulated flanking bands, centered on 800 and 1200 Hz. In the region where hearing

sensitivity was usually relatively poor, the OSB was centered on 2000 Hz. There were two additional conditions in the 2000 Hz region where comodulated flanking bands were present: one using a relatively narrow frequency separation and the other using a wider frequency separation. For the narrow frequency separation, the flanking bands were centered on 1750 and 2250 Hz; for the wider frequency separation, the flanking bands were centered on 1500 and 2500 Hz. The rationale was that poor frequency selectivity might prohibit a large CMR for the narrow frequency separation, but that CMRs might be more normal for the wider frequency separation. Audiograms for four listeners with noise-induced hearing loss are shown in Table 20-1.

Figure 20-1 shows CMR data obtained for the 1000 Hz signal, and Figure 20-2 shows data obtained for the 2000 Hz signal. Each figure shows thresholds obtained for the OSB alone, and with comodulated flanking bands present. The CMR is the difference between these two thresholds. The circles in the figures show average data from three normal-hearing listeners. At 1000 Hz (Figure 20-1) these control listeners exhibited approximately 13 dB of masking release when the flanking bands were present. For the listeners with noise-induced loss, there was also substantial masking release, ranging from about 9 to 11 dB. However, for all four of the listeners with noise-induced hearing loss, the thresholds with the flanking bands present were higher than normal; furthermore, even the baseline (OSB) threshold was substantially higher than normal for subject CC.

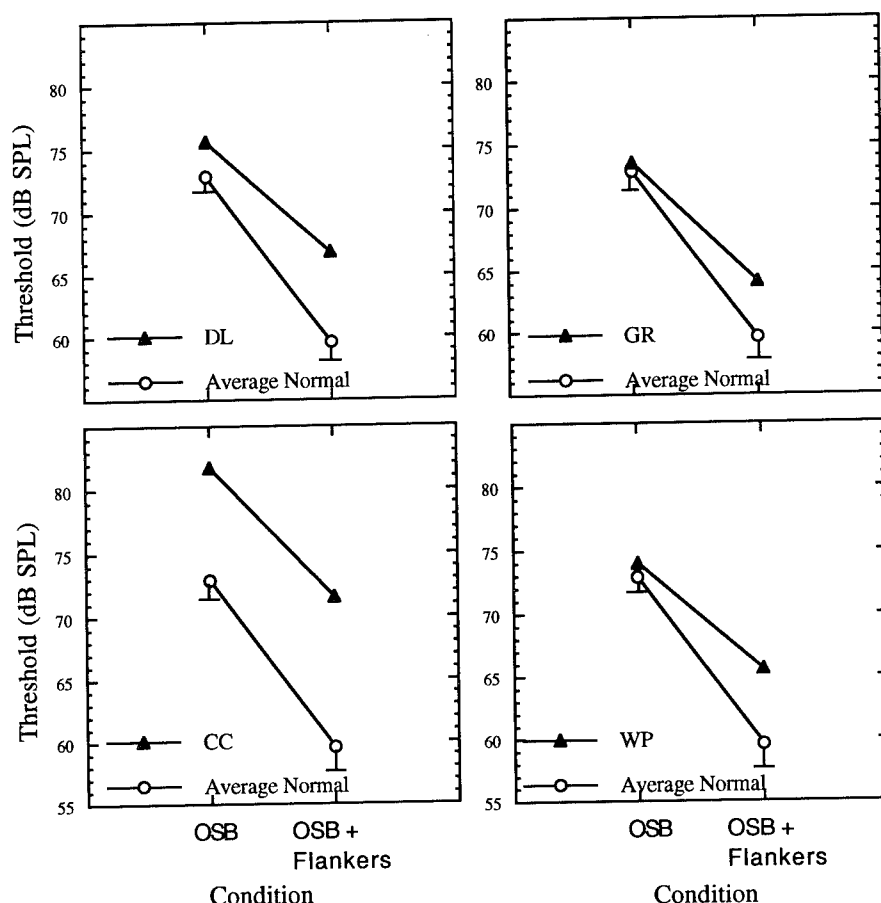


Figure 20-1 1000 Hz signal thresholds for the on-signal band (OSB) condition and the OSB plus flanking band condition (OSB + Flankers). Error bars showing 1 standard deviation are drawn on only one side of the data symbol for the sake of clarity.

Figure 20-2 shows CMR data for the 2000 Hz signal. In addition to the OSB threshold, thresholds with flanking bands present are shown for wide separation and narrow separation. Again, the normal-hearing listeners show appreciable masking release when flanking bands are present, about 9 dB for the wide frequency separation, and about 13 dB for the narrow frequency separation. There were relatively large individual differences among the hearing-impaired listeners. All subjects, except for DL, showed substantial CMR for the wide frequency separation, and subject GR showed substantial CMR both for the wide and the narrow frequency separation. However, three of the four listeners with noise-induced hearing loss showed abnor-

mally small CMRs for the narrow frequency spacing. This result is possibly due to relatively poor frequency selectivity not allowing independent analysis of the closely spaced bands.

In general, the preliminary results on subjects having noise-induced hearing loss agree with results from past studies that employed subjects that were relatively unselected for etiology. That is, the preliminary data on subjects with noise-induced hearing loss showed reduced CMRs in regions of threshold loss, with a pattern of results that is consistent with poor frequency selectivity.

One specific CMR paradigm²² may be strongly related to auditory processes that allow us to process competing sound sources

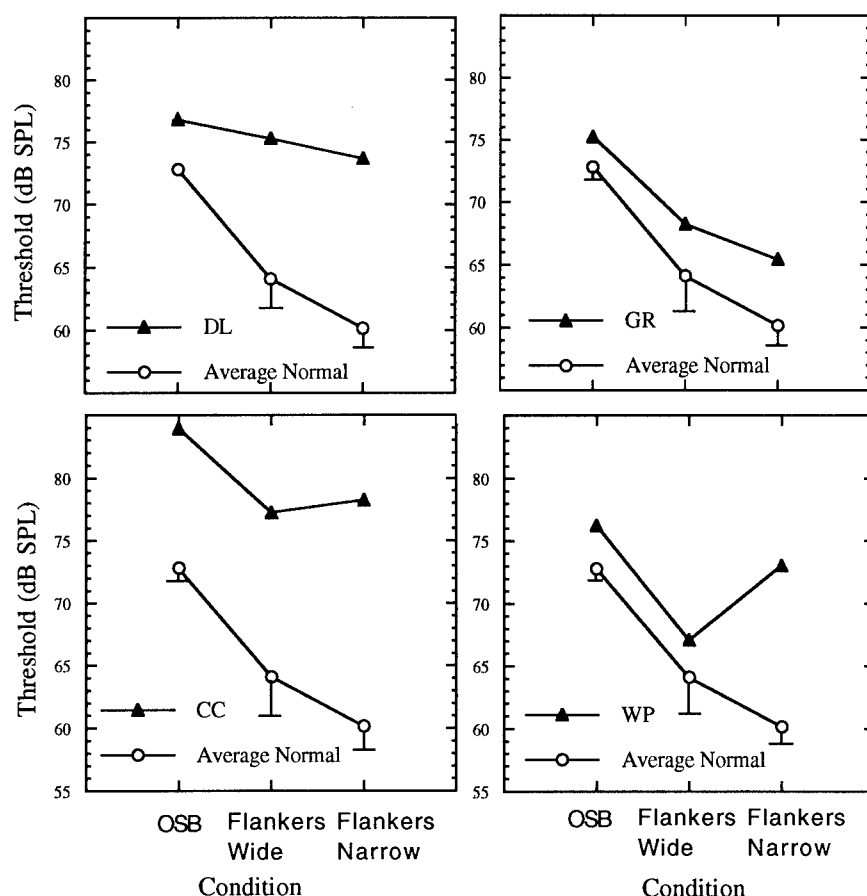


Figure 20-2 2000 Hz signal thresholds for the on-signal band (OSB) condition, and conditions where the flanking bands had wide spectral spacing (Flankers Wide) and narrow spectral spacing (Flankers Narrow). Error bars showing 1 standard deviation are drawn on only one side of the data symbol for the sake of clarity.

that are present simultaneously. In the baseline condition of this paradigm, CMR is determined by contrasting detection of a pure-tone signal when only an on-signal noise band was present with signal detection when several comodulated flanking bands were also present. The comodulated flanking bands generally result in a CMR of between 10 and 15 dB. In another condition, two additional noise bands are added on either side of the OSB. These bands are comodulated with respect to each other, but not with respect to the band centered on the signal. These interposed bands (termed codeviant band) greatly reduce the CMR. One reason for the reduction in CMR is that the codeviant bands may be in-

cluded in across-frequency comparisons with the OSB, resulting in false-positive across-frequency difference cues. The problem may be one of processing multiple auditory sources, one defined by the common modulation on the signal band and its comodulated flanking bands, and the other defined by the codeviant bands. With only the two codeviant bands present, the auditory system may not have enough information to decide whether to group the OSB with its comodulated flanking bands, or with the spectrally proximal codeviant bands.

Hall and Grose²² found that the codeviant bands could be made less disruptive if stimuli were manipulated such that it became more

clear that there were two separate apparent auditory sources. This was accomplished by systematically increasing the number of codeviant bands, while holding the number of bands that were comodulated with the OSB constant. It was found that CMR actually increased as more codeviant bands were added.

It is of interest to apply this paradigm to cochlear hearing-impaired listeners because of the apparent difficulty such listeners report in analyzing competing sounds in the real world. When this paradigm is applied to cochlear-impaired listeners, the results indeed indicate that the presence of codeviant bands are more disruptive for cochlear-impaired listeners than for the normal-hearing listeners.²³ Furthermore, the hearing-impaired listeners show a smaller increase (in dB) of CMR with increasing number of codeviant bands than do normal-hearing listeners. Interestingly, however, the hearing-impaired listeners do not show an abnormally small benefit of increased number of codeviant bands when results are expressed in terms of percent reduction in CMR with codeviant bands present.

It is possible that poor frequency analysis in the cochlear-impaired listeners may contribute to the finding that codeviant bands have a particularly deleterious effect for such listeners. That is, with relatively poor frequency selectivity, it is possible that energy from the codeviant bands leaks into the passbands of the filters processing the comodulated bands. Such leakage would modify the modulation patterns present across the outputs of the auditory filters, therefore reducing CMR. This idea receives support from the finding that the deleterious effect of the proximal codeviant bands is more similar to normal when the overall spacing between noise bands is increased.²³

In summary, there is evidence that masking release associated with comodulation is reduced in listeners with cochlear hearing impairment, and that the masking release is more deleteriously affected in impaired than normal-hearing listeners when codeviant bands are present. There is some reason to believe that these results are linked to de-

graded frequency selectivity. Other factors that may result in reduced CMRs in cochlear-impaired listeners include reduced audibility of stimuli and low SL.

Monaural Envelope Correlation Perception

Another psychoacoustical ability that has been linked to auditory grouping/segregation by common modulation is monaural envelope correlation perception. Richards²⁴ observed that listeners were sensitive to the correlation in the envelope between two narrowband noise stimuli, even when the stimuli were relatively widely separated. Although previous researchers had investigated similar conditions,^{25,26} Richards was the first to show that across-frequency envelope discrimination abilities were relatively robust across a range of conditions in normal-hearing listeners. Although there is not complete agreement across all current studies, monaural envelope correlation discrimination would appear to be facilitated by several stimulus features. For example, performance is generally best for relatively narrow frequency spacings between noise bands,²⁴ relatively wide noise bands,²⁷ monaural as opposed to dichotic presentation of noise bands,²⁸ multiple noise bands,²⁹ and long-duration stimuli.²⁷

It has been speculated that one cue that the auditory system uses to link different frequency components to a single auditory source is across-frequency correlation of amplitude envelopes. Monaural envelope correlation perception paradigms may therefore be regarded as a means of objectively studying this contribution to auditory sound source determination. Because many listeners with cochlear hearing loss complain of difficulty in hearing target sounds in competing noise backgrounds, the investigation of monaural envelope correlation perception in these listeners is important in accounting for part of their auditory disability.

Hall and Grose²⁹ investigated monaural envelope correlation perception in listeners with cochlear hearing loss, investigating factors of the Δf between noise bands and the number

of noise bands present (two, three, or five). Although there were individual differences among the impaired listeners, performance was likely to be reduced in the impaired listeners only when Δf 's were relatively narrow. When Δf 's were 500 Hz or more, the normal-hearing and hearing-impaired listeners had similar performance.

Because the subjects in the study of Hall and Grose²⁹ were relatively unselected in terms of hearing loss etiology, we have since run further monaural envelope correlation perception conditions on a set of four listeners with noise-induced hearing loss. As for the CMR study (above), we examined a 2000 Hz stimulus region, and a 1000 Hz region. We used

pure-tone carriers that were amplitude modulated by 50 Hz low-pass noises. The average depth of modulation was 0.65. For the 1000 Hz region, the carriers were centered on 800, 1000, and 1200 Hz. For the 2000 Hz region there were again two frequency spacings. In one, the carriers were 1750, 2000, and 2250 Hz; in the condition using wider spacing, the carriers were presented at 1500, 2000, and 2500 Hz.

Interestingly, performance for all four of the subjects was relatively poor, even in the frequency region where hearing was relatively normal (see Figure 20-3). Whereas the normal-hearing listeners showed average performance better than 85% correct for all three

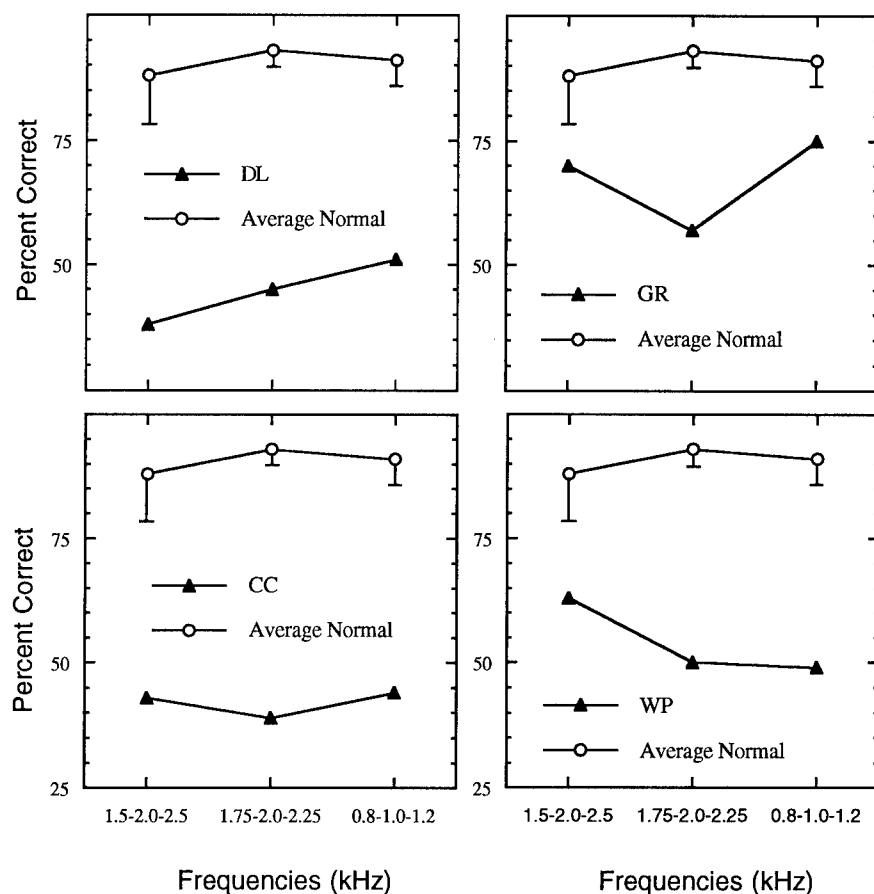


Figure 20-3 Monaural envelope correlation results (Percent Correct) for the 2000 Hz conditions where noise bands were centered on 1500–2000–2500 Hz or 1750–2000–2250 Hz, and the 1000 Hz condition where noise bands were centered on 800–1000–1200 Hz. Error bars showing 1 standard deviation are drawn on only one side of the data symbol for the sake of clarity.

conditions, the performance for the listeners having noise-induced hearing loss was usually substantially worse than this. There was perhaps a slight trend for the listeners with noise-induced hearing loss to have better performance for the 1000 Hz region, but this trend was not very consistent across listeners. The relatively poor performance among the listeners with noise-induced hearing loss is intriguing, and we plan further experiments to examine whether noise-induced hearing loss may be particularly associated with a decrease in the ability to detect correlation of temporal modulation across frequency.

It is interesting to speculate what the ramifications of these results may be for hearing performance in everyday environments. Many important complex sounds have components separated by less than 500 Hz, and thus in the region where impaired listeners show poor performance in detecting envelope correlation. This may mean that one cue for the synthesis of a sound source (coherence of envelope across frequency) is lost to some listeners with cochlear hearing impairments. To effect the segregation of one sound source from another, then, the listener may have to rely on other cues that may be better preserved. One factor that may ameliorate this situation is that for narrow Δf , many cochlear-impaired listeners show a greater than normal boost in performance with increasing number of bands present (due largely to the fact that normal-hearing listeners exhibit near perfect performance with only two bands present). Thus, in situations where information is present and audible across a wide range of frequencies, performance of cochlear-impaired listeners may not suffer greatly.

It is possible that the poor results found for relatively narrow Δf can be blamed upon poor peripheral frequency analysis. Poor frequency selectivity will result in the information at the output of auditory filters centered on the noise bands being less independent: in the limit, a lack of frequency resolution would result in identical envelopes at the outputs of two different auditory filters, regardless of whether the envelopes of the physical noise bands were correlated or uncorrelated. Under this

circumstance, across-channel cues would not be available. Reduced, rather than absent, frequency selectivity would result in a reduction of across-channel cues, particularly for narrow Δf .

Conclusions

Listeners with noise-induced and other forms of cochlear hearing loss show reduced abilities in some psychoacoustic tasks involving the analysis of temporally coherent information that is spread across frequency. Such tasks include CMR and monaural envelope correlation perception. It is possible that the auditory impairments demonstrated through these basic psychoacoustic tasks may contribute to poor analysis of many types of complex signals encountered in real environments. At least part of the decrease in performance on these across-frequency processes is probably attributable to reduced frequency selectivity.

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Chapter 21

Effects of Noise-Induced Hearing Loss on Temporal Resolution

Brian C.J. Moore

Although noise-induced hearing loss produces a characteristic pattern of hearing loss as a function of frequency, its effects on temporal processing seem to be similar to those produced by other hearing losses that are primarily of cochlear origin.^{1,2} Because many studies of temporal processing in subjects with cochlear hearing loss have not clearly identified the origin of the loss (noise induced as opposed to other forms of cochlear damage), this chapter considers data from subjects with cochlear damage caused in a variety of ways.

Many studies have shown that temporal resolution can be adversely affected by cochlear hearing loss. At first sight, it is not obvious why this should be the case. Neurophysiological data obtained from animals with noise-induced or other cochlear damage indicate that the temporal patterns of supra-threshold stimuli are accurately represented in the responses of primary auditory neurones. Furthermore, the auditory filters in hearing-impaired subjects are typically broader than normal,³⁻⁵ which might be expected to lead to better than normal temporal resolution. Nevertheless, psychoacoustic data indicate that temporal processing for certain types of stimuli is strongly affected by cochlear hearing loss, with temporal resolution usually being worse than normal. To understand the reasons for these effects, it is helpful to use a model of temporal processing in the normal auditory system and to consider how the different stages of the model may be altered by cochlear pathology.

In characterizing temporal resolution in the auditory system, it is essential to take account of the filtering that takes place in the peripheral auditory system. Temporal resolution depends on two main processes: analysis of the time pattern occurring within each frequency channel and comparison of the time patterns across channels. This chapter concentrates mainly on within-channel processes, because there have been few studies of across-channel processing in hearing-impaired subjects.

Modeling Within-Channel Temporal Resolution

Models of temporal resolution assume that the internal representation of stimuli is "smoothed" over time, so that rapid temporal changes are lost but slower ones are preserved. Several models of temporal resolution have the general form illustrated in Figure 21-1. There is an initial stage of bandpass filtering, reflecting the action of the auditory filters. For simplicity, only one filter is shown; in reality there would be an array of parallel channels, each like that shown in the figure. Each filter is followed by a nonlinear device. The output of the nonlinear device is fed through a "smoothing" device that can be implemented either as a low-pass filter⁶ or a sliding temporal integrator.⁷ The output of the smoothing device is fed to a decision device.

Effect of Auditory Filtering

When a signal is passed through a bandpass filter, the time pattern of the signal is smeared

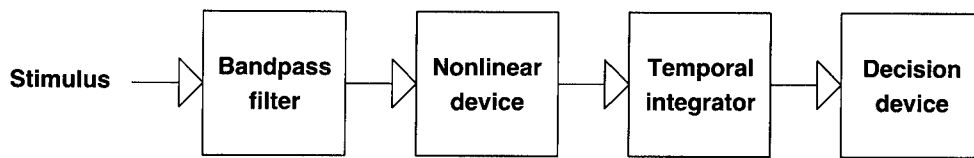


Figure 21-1 A block diagram showing the stages typically found in models of temporal processing.

or smoothed somewhat by the filter. Generally, the narrower the filter, the greater the smoothing. The auditory filters have bandwidths that decrease progressively with decreasing center frequency.⁸ One might expect, therefore, that the auditory filters would play some role in limiting temporal resolution, this effect being greater at low center frequencies.

Many studies have addressed the question of whether temporal resolution does vary with center frequency. Green⁹ used time-reversed stimuli where each stimulus consisted of a brief pulse of a sinusoid in which the level of the first half of the pulse was 10 dB different from that of the second half. Subjects were required to distinguish two signals, differing in whether the half with the high level was first or second. Green measured performance as a function of the total duration of the stimuli. The threshold was similar for center frequencies of 2 and 4 kHz, and was between 1 and 2 milliseconds. However, the threshold was slightly higher for a center frequency of 1 kHz, being between 2 and 4 milliseconds.

Several researchers have measured thresholds for detecting gaps in narrowband noises.¹⁰⁻¹² When a temporal gap is introduced into a narrowband sound, energy "splatter" occurs outside the nominal frequency range of the sound. To prevent the splatter being detected, the sounds are presented in a background sound, usually a noise, designed to mask the splatter. Gap thresholds have often been found to decrease monotonically with increasing center frequency. However, in these experiments, the bandwidth of the stimuli increased with increasing center frequency. Noise bands have inherent fluctuations in amplitude, and the rapidity of these fluctuations increases with increasing bandwidth. Gap thresholds for noise bands

may be partly limited by the inherent fluctuations in the noise.^{11,13,14} Randomly occurring dips in the noise may be "confused" with the gap to be detected. The confusion would be maximal for dips comparable in duration to the gap. In practice, this means that noise with a narrow bandwidth, and hence slow fluctuations, would create the greatest confusion and give the largest gap thresholds. The data are consistent with this view: gap thresholds for narrowband noises increase with decreasing noise bandwidth.^{11,13,15,16} Furthermore, gap thresholds measured using noise bands of fixed width show little effect of center frequency.^{13,15,17}

Shailer and Moore¹⁸ studied the ability of subjects to detect a temporal gap in a sinusoid. To mask splatter associated with the introduction of the gap, the sinusoid was presented in a continuous noise with a spectral notch at the frequency of the sinusoid. The results were strongly affected by the phase at which the sinusoid was turned off and on to produce the gap. Initially, only the simplest case will be considered here, called "preserved phase" by Shailer and Moore. In this case the sinusoid was turned off at a positive-going zero crossing (i.e., as the waveform was about to change from negative to positive values) and it started (at the end of the gap) at the phase it would have had if it had continued without interruption. Thus, for the preserved-phase condition it was as if the gap had been "cut out" from a continuous sinusoid. For this condition, the detectability of the gap increased monotonically with increasing gap duration.

Shailer and Moore¹⁸ found that the gap threshold was roughly constant at about 5 milliseconds for center frequencies of 400, 1000, and 2000 Hz. Recently, Moore et al.¹⁹ measured gap thresholds for center frequencies of

100, 200, 400, 800, 1000, and 2000 Hz, using a condition similar to the preserved-phase condition of Shailer and Moore. The gap thresholds were almost constant, at 6–8 milliseconds over the frequency range 400–2000 Hz, but increased somewhat at 200 Hz, and increased markedly, to about 18 milliseconds, at 100 Hz. Individual variability also increased markedly at 100 Hz.

Overall, the results of experiments using narrowband stimuli indicate that temporal resolution does not vary markedly with frequency, except at frequencies of 200 Hz and below. This suggests in turn that the smoothing produced by the auditory filters does not play a major role, except at very low frequencies.

Two of the conditions in the experiment of Shailer and Moore¹⁸ did, however, show effects that could be attributed to “ringing” in the auditory filter. In the condition that they called “standard phase” the signal was turned off (to start the gap) at a positive-going zero crossing, and turned on again (at the end of the gap) at a positive-going zero crossing. An example of a psychometric function for the standard-phase condition is shown in Figure 21-2. The frequency of the sinusoid was 400 Hz, so its period, P , was 2.5 milliseconds. A two-alternative forced-choice task was used, so chance performance corresponds to 50% correct. The psychometric function is distinctly nonmonotonic. The gap is difficult to detect when its value is an integer multiple of P , that is, 2.5 and 5 milliseconds. Conversely, the gap is easy to detect when its value is $(n + 0.5)P$, where $n = 0$ or 1. Nonmonotonic psychometric functions were also found for a frequency of 1000 Hz but not for a frequency of 2000 Hz.

Shailer and Moore¹⁸ explained these results in terms of ringing in the auditory filter. Their argument is illustrated in Figure 21-3, which shows responses of a simulated auditory filter with a center frequency of 400 Hz to a series of stimuli from the standard-phase condition, with gap durations ranging from 1.2 to 3.7 milliseconds. When the sinusoid is turned off at the start of the gap, the filter continues to respond or ring for a certain time. If the gap

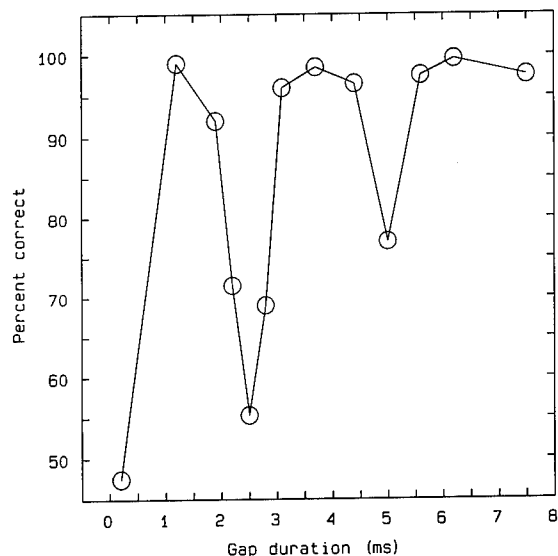


Figure 21-2 The percentage correct detection of a gap in a 400 Hz sinusoid, plotted as a function of the gap duration. The gap started at a positive-going zero crossing of the sinusoid and the sinusoid started after the gap at a positive-going zero crossing, a condition referred to as standard phase.

duration is 2.5 milliseconds, corresponding to one whole period of the sinusoid, the sinusoid following the gap is in phase with the ringing response. In this case the output of the filter shows only a small dip, and we would expect gap detection to be difficult. This is exactly what is observed. For a gap duration of 1.2 or 3.7 milliseconds, the sinusoid following the gap is out of phase with the ringing response. Now the output of the filter passes through zero before returning to its steady-state value. The resulting dip in the filter output is larger, and is much easier to detect. This explains why the psychometric function is nonmonotonic for the standard-phase condition. For the preserved phase condition, the sinusoid following the gap is always in phase with the ringing response of the auditory filter. Thus, the dip in the output of the auditory filter increases monotonically with increasing gap duration, and the psychometric function is monotonic. The psychometric function at high frequencies is always monotonic, even for the standard-phase condition, because the

auditory filter is sufficiently broad that ringing lasts only for a very short time.

In summary, the auditory filter can have a strong influence on the pattern of results for deterministic stimuli, especially at low frequencies. However, it appears to play only a minor role in limiting temporal resolution, except perhaps at very low frequencies.

Characteristics of Nonlinearity and Smoothing Device

Although all the stages of the model can affect temporal resolution, it is often felt that the stage that is most directly related to temporal resolution is the low-pass filter or temporal integrator. Thus, it would be desirable to be able to determine the characteristics of this stage independently of the other stages. In practice, this is very difficult to do.²⁰

Some recent experiments^{7,21} have attempted to determine the characteristics of the temporal integrator on the assumption that the nonlinearity has a square-law characteristic. In other words, the temporal integrator was assumed to act on a powerlike quantity. The integrator itself was modeled as a weighting function, or "window," that performs a running average of the power at the output of the auditory filter. The experimental data used to derive the window shape were thresholds for detection of a very brief tone pulse presented between two bursts of noise, measured as a function of the time interval from the end of the first burst to the center of the tone pulse, and the center of the tone pulse to the start of the second noise burst (combined forward and backward masking).

Some examples of temporal window shapes derived in this way are shown in Figure 21-4, taken from Plack and Moore.²¹ In these examples, each side of the temporal window was modeled as the sum of two rounded exponential functions. The shape of the temporal window is almost invariant with center frequency, except for a slight broadening (associated with poorer temporal resolution) at low center frequencies. The temporal window broadens somewhat as the sound level decreases. A

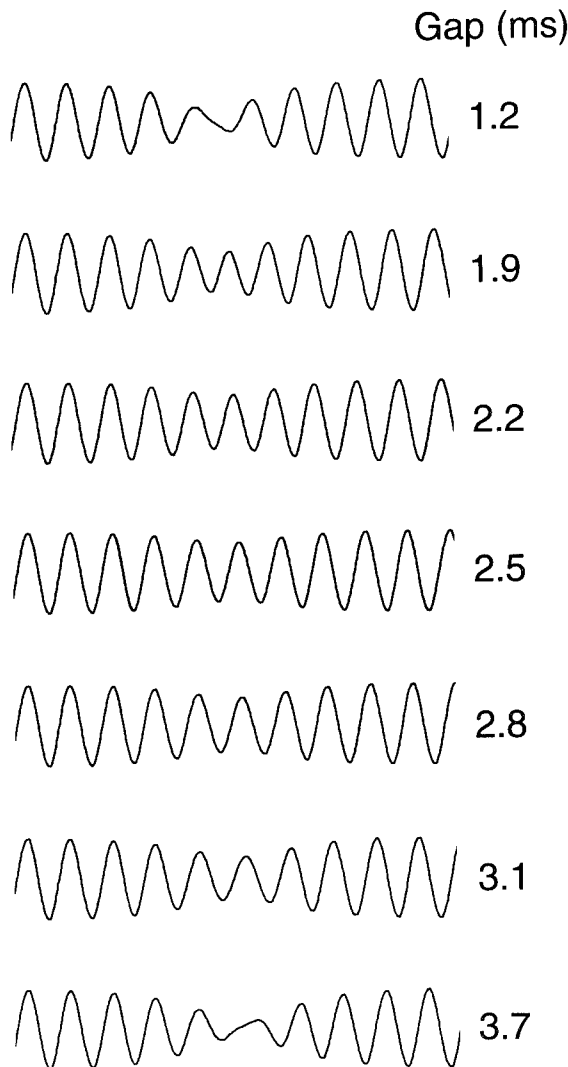


Figure 21-3 Simulation of the output of an auditory filter centered at 400 Hz in response to stimuli in the preserved-phase condition. The duration of the gap is indicated for each waveform.

simple summary measure of the temporal window is its equivalent rectangular duration (ERD). The ERDs found by Plack and Moore were typically about 8–9 milliseconds.

One problem with this model is that it does not correctly account for the way that backward and forward masking combine. If a backward masker and a forward masker are equated for effectiveness when each is presented separately, and then the two maskers are com-

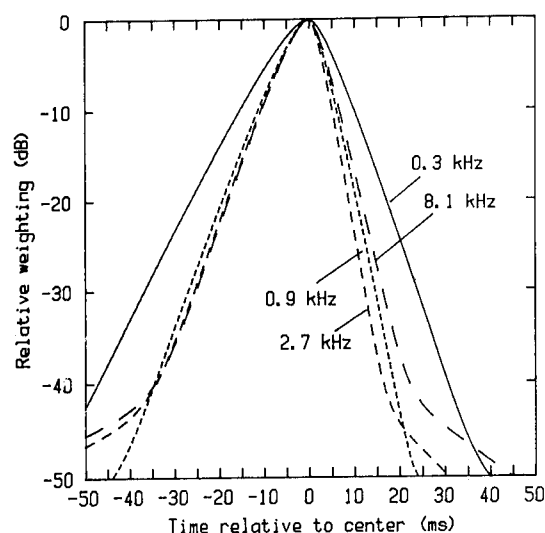


Figure 21-4 Temporal window shapes derived by Plack and Moore²¹ for normally hearing subjects. Each curve represents a different center frequency.

bined, the resulting amount of masking is usually greater than would be predicted from a simple linear summation of the effects of the individual maskers (a 3 dB increase).^{22,23} To account for this effect, several workers have suggested models of temporal resolution in which there is a compressive nonlinearity prior to the temporal integrator.^{23–26} Oxenham and Moore²³ measured forward and backward masking separately and in combination for a very brief 6 kHz signal and a noise masker. They showed that the results could be used to separate the effects of the temporal integrator and the nonlinearity prior to the integrator. The results were fitted best by a nonlinearity where the rectified amplitude was raised to a power between 0.5 and 0.7. The inclusion of a compressive nonlinearity in the model has a marked effect on the derived temporal window. Specifically, the more compressive the nonlinearity, the broader the temporal window.^{24,26} Oxenham and Moore, using rounded exponentials to model the window shape, found ERDs of about 12–15 milliseconds. When the window was modeled by exponential functions, the ERDs were smaller, typically about 8–10 milliseconds.

These values are broadly consistent with ERDs estimated in other ways.²⁶

The physiological correlate of the compressive nonlinearity in the model described above is still somewhat uncertain. However, it can probably be partly related to the compressive nonlinearity that is observed in input–output functions of the basilar membrane in the normal cochlea.^{27–29} This compressive nonlinearity depends on the operation of an active mechanism in the cochlea that is extremely susceptible to cochlear damage by noise exposure or other insults³⁰ and probably depends on the integrity of the outer hair cells.

Temporal Resolution in Hearing-Impaired Subjects

Some measures of temporal resolution in subjects with cochlear hearing loss appear to show reduced temporal resolution, while others do not. Several factors can affect the results, and not all of these are directly connected with temporal processing itself.

Influence of Sound Level

One important factor influencing measures of temporal resolution is the sound level used. Many measures of temporal resolution show that performance in normally hearing subjects worsens at low sensation levels (SLs).^{11,12,31,32} This is not unique to temporal resolution; performance on many tasks worsens at low SLs, presumably because less neural information is available, or because of the greater effects of internal noise at low SLs. It is not generally possible to test hearing-impaired subjects at high SLs, because they have loudness recruitment; sounds with levels of 90–100 dB sound pressure level (SPL) appear as loud as they would to a normal listener. Thus, on some measures of temporal resolution, such as the detection of gaps in bands of noise or the rate of recovery from forward masking, hearing-impaired subjects appear markedly worse than normal subjects when tested at the same SPLs, but only slightly worse at equal SLs.^{2,33,34}

For deterministic stimuli that have no inherent random fluctuations, hearing-impaired subjects can actually perform a little better than normally hearing subjects when tested at equal SLs. This applies to the detection of gaps in sinusoids^{35,36} and to the discrimination of Huffman sequences³⁷; the latter are clicklike sounds with identical power spectra but different phase spectra.

Influence of Audible Bandwidth

Another important consideration is the bandwidth available to the listeners. This can be clearly seen by consideration of studies measuring the temporal modulation transfer function (TMTF). The TMTF shows the amount of amplitude modulation required for detection of the modulation, plotted as a function of modulation rate.⁶ It is generally assumed that the ability to detect modulation at high rates is limited by the temporal resolution of the ear. Several studies measuring TMTFs for broadband noise carriers showed that impaired listeners were generally less sensitive to high rates of modulation than normal listeners.³⁸⁻⁴⁰ However, this may have been largely a consequence of the fact that high frequencies were inaudible to the impaired listeners⁴⁰; most of the subjects used had greater hearing losses at high frequencies than at low, as is typical in cases of noise-induced hearing loss. When the broadband noise is low-pass filtered, as a crude simulation of the effects of threshold elevation at high frequencies, normally hearing subjects also show a reduced ability to detect modulation at high rates.⁴⁰

Bacon and Gleitman⁴¹ measured TMTFs for broadband noise using subjects with relatively flat hearing losses. They found that at equal (high) SPLs performance was similar for hearing-impaired and normally hearing subjects. At equal (low) SLs, the hearing-impaired subjects tended to perform better than the normally hearing subjects. Moore et al.⁴² controlled for the effects of listening bandwidth by measuring TMTFs for an octave-wide noise band centered at 2 kHz, using subjects with unilateral and bilateral cochlear hearing loss.

Over the frequency range covered by the noise, the subjects had reasonably constant thresholds as a function of frequency, both in their normal and their impaired ears. This ensured that there were no differences between subjects or ears in terms of the range of audible frequencies in the noise. To ensure that subjects were not making use of information from frequencies outside the nominal pass-band of the noise, the modulated carrier was presented in an unmodulated broadband noise background. The results for the subjects with unilateral impairments are shown in Figure 21-5. It can be seen that performance is similar for the normal and impaired ears, both at equal SPL and equal SL, although there is a slight trend for the impaired ears to perform better at equal SL.

Studies of gap detection also show clear effects of the audible frequency range of the stimuli. For a broadband noise marker, gap thresholds become progressively larger as the audible frequency range of the stimuli is reduced by increasing high-frequency hearing loss.^{12,43,44}

To summarize the results so far, subjects with cochlear hearing loss often show reduced temporal resolution as a result of the low SL of the stimuli and/or the reduced audible bandwidth of the stimuli. When these factors are controlled for, hearing-impaired subjects often perform as well as, or even better than normal.

Influence of Broadened Auditory Filters

Subjects with cochlear hearing loss usually have auditory filters that are broader than normal.³⁻⁵ As mentioned earlier, one might expect that this would lead to improved temporal resolution. However, because the auditory filters in normal ears appear to play little role in limiting temporal resolution, except at very low frequencies, it has proved difficult to demonstrate changes in temporal resolution resulting from broadened auditory filters.

One case where changes attributable to the auditory filters have been found is for the detection of temporal gaps in sinusoids. As de-

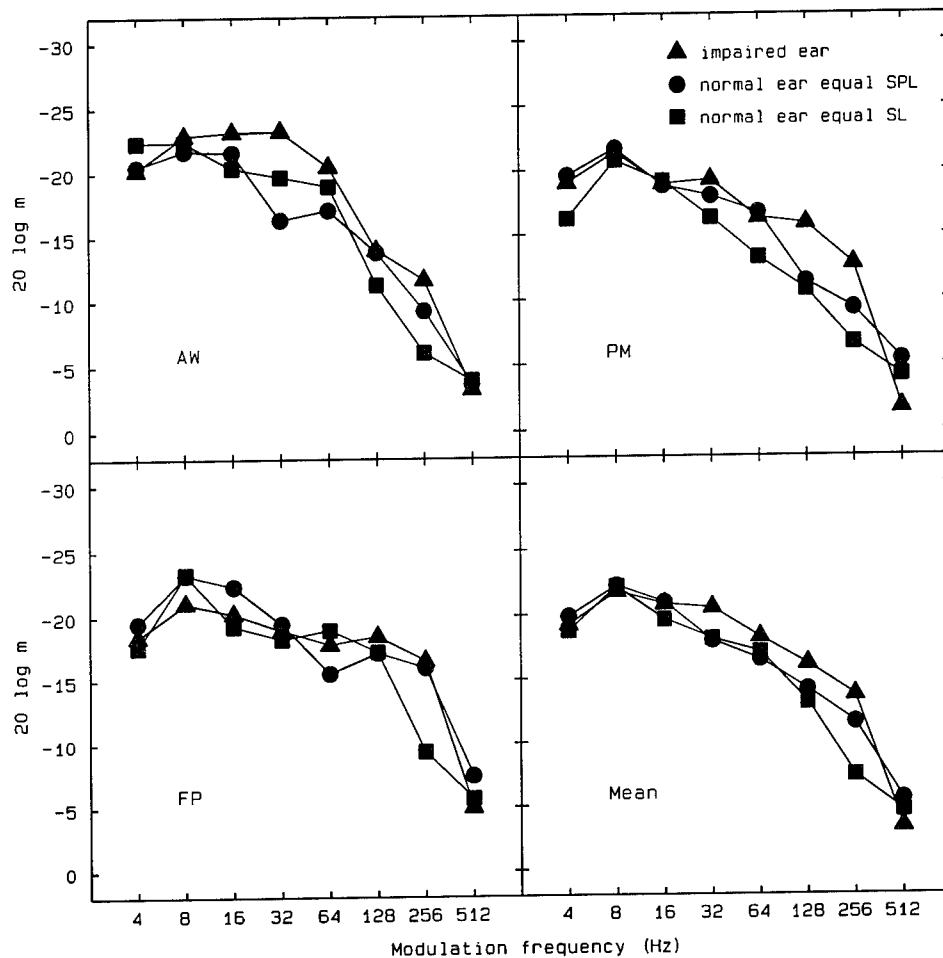


Figure 21-5 Temporal modulation transfer functions (TMTFs) obtained using a bandpass noise carrier for the normal and impaired ears of three subjects with unilateral cochlear hearing loss.

scribed earlier, for the “preserved phase” condition of Shailer and Moore,¹⁸ psychometric functions for normal ears at low frequencies are nonmonotonic (see Figure 21-2). This was explained in terms of ringing in the auditory filters. For subjects with cochlear hearing loss, the psychometric functions are monotonic.³⁶ Moore et al.³⁶ showed that this could be explained by the faster temporal response of the auditory filters that resulted in a large dip in the filter output even when the gap corresponded to a small whole number of periods. Thus, the psychometric functions for impaired ears resemble those of normal ears at high frequencies (where the auditory filters are also broad); in both cases, performance

appears to be limited by the more central temporal integration process.

Influence of Changes in Compressive Nonlinearity

For certain types of sounds, the temporal resolution of subjects with cochlear hearing loss seems to be worse than normal even when the stimuli are well above threshold and when all of the components of the stimuli fall within the audible range. This happens mainly for stimuli that contain slow random fluctuations in amplitude, such as narrow bands of noise. For such stimuli, subjects with cochlear impairment often perform more poorly than nor-

mal in tasks such as gap detection.^{12,33,34,43} However, gap detection is not usually worse than normal when the stimuli are sinusoids, which do not have inherent amplitude fluctuations.^{35,36} Glasberg et al.³⁴ and Moore and Glasberg³⁵ suggested that the poor gap detection for narrowband noise stimuli might be a consequence of loudness recruitment, the abnormally rapid growth of loudness with increasing intensity that occurs commonly in cases of cochlear hearing loss, including noise-induced loss.⁴⁵ For a person with recruitment, the inherent fluctuations in a narrowband noise would result in larger than normal loudness fluctuations from moment to moment, so that inherent dips in the noise might be more confusable with the gap to be detected.

This idea can also be expressed in terms of the model of temporal resolution described earlier. It seems likely that loudness recruitment is caused primarily by a reduction in or loss of the compressive nonlinearity found in the normal cochlea.⁴⁶⁻⁴⁹ When cochlear damage occurs, the cochlea behaves in a more linear way, and the input-output function of the basilar membrane becomes less compressive, having a slope closer to unity (on log-log coordinates).

To assess this idea, Glasberg and Moore¹⁶ processed the envelopes of narrow bands of noise to modify the envelope fluctuations. The envelope was processed by raising it to a power, N . For stimuli of constant amplitude, for example sinusoidal tones, the level (in decibels) of stimuli processed in this way is a linear function of the level of the unprocessed stimuli, with slope N . This reproduces one of the features of loudness recruitment seen in subjects with unilateral hearing loss; for levels below about 90–100 dB SPL, the level of sound needed in the normal ear to match the loudness of a sound in the impaired ear is roughly a linear function of the level in the impaired ear.

If N is greater than unity, this has the effect of magnifying fluctuations in the envelope, thus simulating the effects of recruitment; higher powers correspond to greater degrees of simulated recruitment. If N is less than

unity, fluctuations in the envelope are reduced. This represents a type of processing that might be used to compensate for recruitment; it resembles the operation of a fast-acting compressor or automatic gain control system.

Values of N used were 0.5, 0.66, 1.0, 1.5, and 2. For $N = 1$, the stimuli were the same as unprocessed Gaussian noise. A value of $N = 2$ simulates the type of recruitment typically found in cases of moderate to severe sensorineural loss, where, for example, a 50 dB range of stimulus levels gives the same range of loudness as a 100 dB range of stimulus levels in a normal ear.⁴⁶

Some examples of the envelopes of unprocessed and processed stimuli are shown in Figure 21-6. The envelopes are plotted on a logarithmic (decibels) scale, because this seems more relevant to loudness perception than a linear amplitude scale. The bottom panel shows the envelope of a "normal" noise band ($N = 1$) with a bandwidth of 10 Hz. The top panel shows the effect of squaring the envelope ($N = 2$), and the middle panel shows the result of raising the envelope to the power

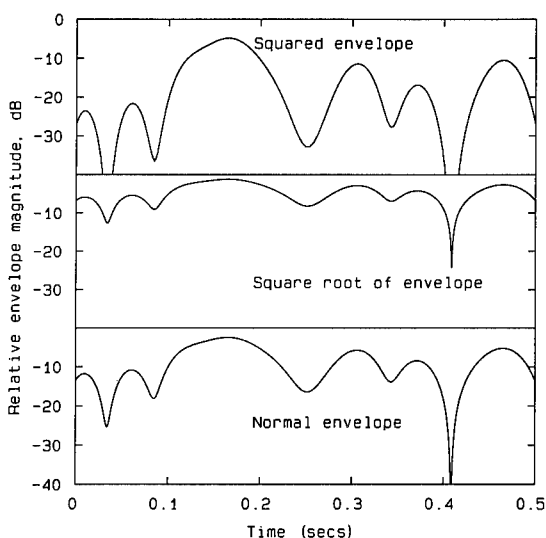


Figure 21-6 Examples of the envelopes of noise bands with $N = 1$ (unprocessed, bottom panel), $N = 0.5$ (middle panel), and $N = 2$ (top panel). The noise bandwidth was 10 Hz. The envelope magnitudes are plotted on a decibel scale.

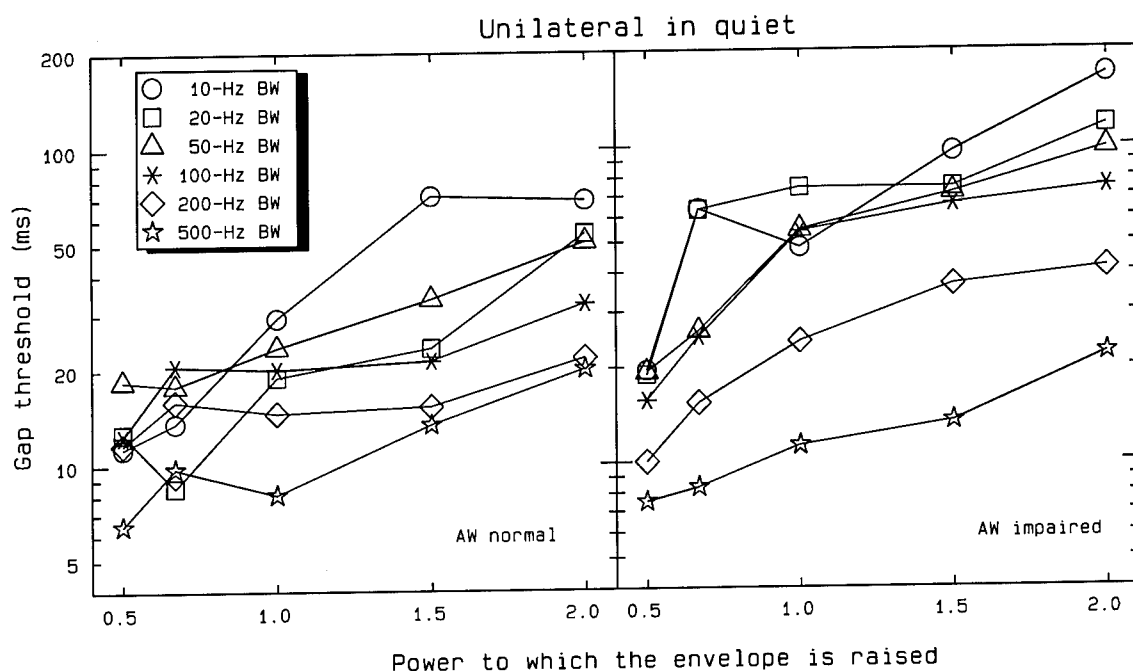


Figure 21-7 Thresholds for detecting a gap in a noise band whose envelope had been processed to enhance or reduce fluctuations. Gap thresholds are plotted as a function of the power to which the envelope was raised, with noise bandwidth as parameter: the higher the power, the greater the fluctuations. Results are shown for each ear of a subject with unilateral cochlear hearing loss.

0.5. The envelope fluctuations are obviously greatest in the top panel and smallest in the middle panel.

To prevent the detection of spectral splatter associated with the gap or with the envelope processing, the stimuli were presented in a continuous background noise. The spectrum of the noise was chosen so that it would be as effective as possible in masking the splatter while minimizing its overall loudness.

Figure 21-7 shows an example of results obtained using a subject with unilateral hearing loss of cochlear origin. The stimuli were presented at 85 dB SPL, a level well above threshold for both the normal and impaired ears (although the SL was lower in the impaired ear). The results for the normal ear were very similar to those of three normally hearing subjects who were also tested. Gap thresholds increased significantly with decreasing noise bandwidth. This is as expected, because the inherent fluctuations in the noise are slower, and more confusable with the gap to be detected, when the bandwidth is narrow.^{11,50}

For all noise bandwidths, gap thresholds increased as N increased. This effect was particularly marked for the smaller noise bandwidths. There was a significant interaction between bandwidth and N , reflecting the fact that changes in gap threshold with N were greater for small bandwidths. This supports the idea that fluctuations in the noise adversely affect gap detection; greater fluctuations lead to worse performance, especially when the fluctuations are slow.

Gap thresholds were larger for the impaired than for the normal ear. The overall geometric mean gap threshold was 12.8 milliseconds for the normal ear and 27.2 milliseconds for the impaired ear. Performance for the normal ear with $N = 2$ was roughly similar to performance for the impaired ear with unprocessed noise bands ($N = 1$); geometric mean gap thresholds were 26.9 milliseconds for the former and 26.5 milliseconds for the latter. Thus, the simulation of recruitment in the normal ear was sufficient to produce impaired gap detection, comparable to that actually found

in the impaired ear. This is consistent with the dynamic ranges inferred from audiometric data; the range from threshold to discomfort at 1 kHz was 90 dB in the normal ear and 45 dB in the impaired ear.

Reduction of fluctuations, by raising the envelope to a power less than one, produced a small improvement in performance for the normal ear. Values of $N < 1$ gave a marked improvement for the impaired ear. Performance for the impaired ear with the envelope raised to the power 0.5 was comparable to, or even slightly better than, that for the normal ear with unprocessed stimuli; geometric mean thresholds were 11.6 milliseconds for the former and 12.5 milliseconds for the latter. This is again consistent with the fact that the dynamic range in the impaired ear was about half that in the normal ear. Thus, the impaired detection of gaps in noise bands occurring for an impaired ear can be restored to normal by appropriate compression of fluctuations in the envelopes of the stimuli.

For both normal-hearing and hearing-impaired subjects, the effects of changing N decreased with increasing noise bandwidth. One reason for this is that slow fluctuations can be followed by the auditory system, whereas rapid fluctuations are smoothed to some extent by the central temporal integration process described earlier. Rapid fluctuations in the wider noise bands would have been smoothed in this way, thus reducing their influence on gap detection.

The results suggest that, for most hearing-impaired subjects, recruitment, or equivalently, a reduction in the peripheral compressive nonlinearity, may provide a sufficient explanation for increased gap thresholds. Thus, it is not usually necessary to assume any abnormality in temporal processing occurring after the cochlea. However, one subject with a unilateral hearing loss tested by Glasberg and Moore¹⁶ showed higher gap thresholds in the impaired ear than would be expected just on the basis of loudness recruitment. This may have been due to the very low SL of the stimuli for that subject, or to some other factor affecting temporal resolution. Earlier reports have suggested that a few subjects show impair-

ments in temporal resolution even using non-fluctuating stimuli.^{35-37,51} It is possible that the subjects showing this impaired resolution had damage to both outer hair cells (affecting the active process and the compressive nonlinearity) and inner hair cells (affecting the transduction of information), or that they had a retrocochlear component to their hearing loss.

Summary

The overall pattern of results from subjects with cochlear hearing loss can be interpreted in the following way. The central temporal integration mechanism (low-pass filter or sliding temporal window) is probably normal. However, the nonlinearity preceding the integrator is less compressive in ears with a cochlear impairment than in normal ears (reflecting the loudness recruitment in the impaired ears). For stimuli with inherent slow amplitude fluctuations (such as narrow bands of noise) this can lead to poorer gap detection in impaired ears because the inherent fluctuations become more confusable with the gap to be detected. However, for deterministic stimuli (such as sinusoids) or for broadband noise stimuli, it can actually lead to better performance in impaired than in normal ears, when the comparison is made at equal SLs. In practice, hearing-impaired subjects often show poor performance on measures of temporal resolution because the stimuli are at low SLs and/or because the audible bandwidth of the stimuli is restricted.

Acknowledgments

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Chapter 22

Psychoacoustic Performance in Workers With NIHL

Raymond Hétu and Hung Tran Quoc

An Ecological Framework for Performance Impairment Analysis

A considerable amount of data has been gathered on the effects of sensorineural hearing loss, in general, on auditory performance and on noise-induced hearing loss (NIHL), in particular. A major portion of this work characterizes alterations in basic auditory functions, such as hearing sensitivity, frequency resolution, temporal resolution, and loudness function. Partial loss of these functions can be referred to as an "impairment" according to the WHO nomenclature of the effects of chronic disorders.¹ Systematic characterization of impairments now provides an articulate description of the pathological condition and gives clues to help understand the underlying mechanism. However, these impairments as such may be inconsequential if they do not result in reduced auditory capacity to perform specific tasks in everyday life. "Disability" (in the WHO nomenclature) emerges when a dysfunction is actualized, that is, when the individual with an impairment is confronted with a specific demand in a given environment, and consequently experiences a "handicap," that is, a disadvantage in fulfillment of a role that is normal for that individual. Figure 22-1 outlines the relationship between the different effects of hearing damage due to noise. To illustrate, a worker with NIHL can be required to monitor a production line paying attention to a sound warning signal (an auditory demand). If the loss of frequency selectivity (an impairment) resulting from NIHL is

severe enough and if the signal-to-noise ratio is too low considering the capacity of that individual, the latter will experience a disability in that situation; as a result, the worker may be less capable of ensuring his/her safety at the workstation (a handicap).

Performance impairment, in the present account, is explicitly viewed within an ecological perspective that focuses on the *interaction* between individual capacities and environmental demands.² This perspective leads one to analyze performance impairment in terms of compatibility or mismatch between demands and capacities. In other words, it allows one to describe the effects of hearing loss within the environments in which hearing impaired workers evolve. First, hearing impairments associated with NIHL are reviewed briefly. Second, typical auditory demands in the workplace are examined. Third, performance impairments in the work setting are analyzed. Finally, a procedure is proposed to match auditory demands in the workplace with the altered auditory capacities of hearing-impaired workers.

Hearing Impairments Associated With NIHL

Functional limitations in workers with NIHL are relatively well known. They can be summarized as follows:

- *Loss of hearing sensitivity* per se, usually more pronounced in the 3–6 kHz frequency band, is statistically predictable

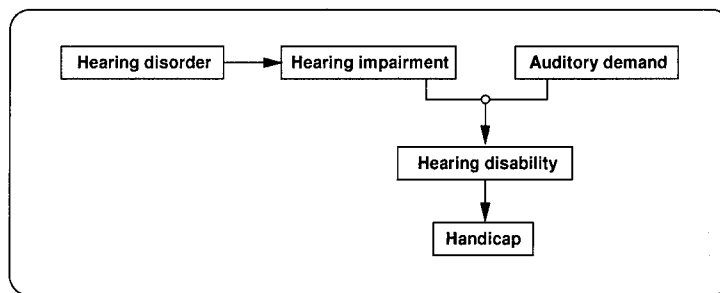


Figure 22-1 Diagram illustrating how a hearing disorder can result in a handicap using the WHO nomenclature.

from the noise exposure level to which workers are subjected.³

- *Loss of frequency resolution* has been quite extensively documented.⁴ This is responsible for the most strongly felt effects of NIHL, namely, the experience of hearing difficulties when there are competing signals.⁵ As illustrated below, loss of frequency selectivity is correlated with loss of sensitivity, although relatively large individual differences are observed at comparable elevated hearing levels.
- *Loss of temporal resolution*⁶ is also correlated with hearing threshold elevation.
- *Compressed loudness function* is observable in the frequency regions where there is a loss of sensitivity.⁷
- *Loss of spatial resolution* may result from the impairments listed above.⁸
- *Persistent annoying tinnitus* is also relatively common among individuals with NIHL,⁹ a condition that may impair concentration.¹⁰

The hearing impairments listed above reflect significant changes in the processing of auditory signals by the inner ear. But actual performance impairment in the work setting depends on the prevailing auditory demands.

Overview of Auditory Demands in Industrial Workplaces

Very few studies have been conducted to characterize specific auditory demands in the industrial work environment. In a recent survey conducted in a large metal products plant of more than 700 workers, close to 100 differ-

ent conditions of use of auditory warning signals were identified.¹¹ The aim of such signals included safety, production, and communication. Lack of distinctiveness was a general feature of most of the signals, the same three types of sources (buzzers, bells, and sirens) being used for a variety of functions. They were superimposed over background noise levels ranging from 67 to 106 dBA with an average of 89.6 dBA (± 9.4 SD). More than one-third of the signals did not meet the level required for them to be recognizable. One out of six signals was much too loud considering the background noise. These findings are but one example of what may be found in industrial settings, where so little is done to optimize the effectiveness of auditory warning signals.

A framework has recently been proposed to characterize auditory demands and match them with auditory capacities in industry.¹¹ It is reproduced in Figure 22-2. The various types of tasks that are performed, whether they involve detection, recognition, or localization, call on different specific capacities depending on the exact nature of the sound signal's action on the auditory system. The acoustic parameters of these signals is further determined by the type of sound sources, and also by their distinctiveness. Auditory demands are further governed by the propagating environment and, in some instances, by the receptive devices used.

Typically, industrial settings are characterized by relatively high levels of background noise, that is, very often close to or above 85 dBA,¹² with reverberation times of 3–5 seconds.¹³ Detection, discrimination, identifica-

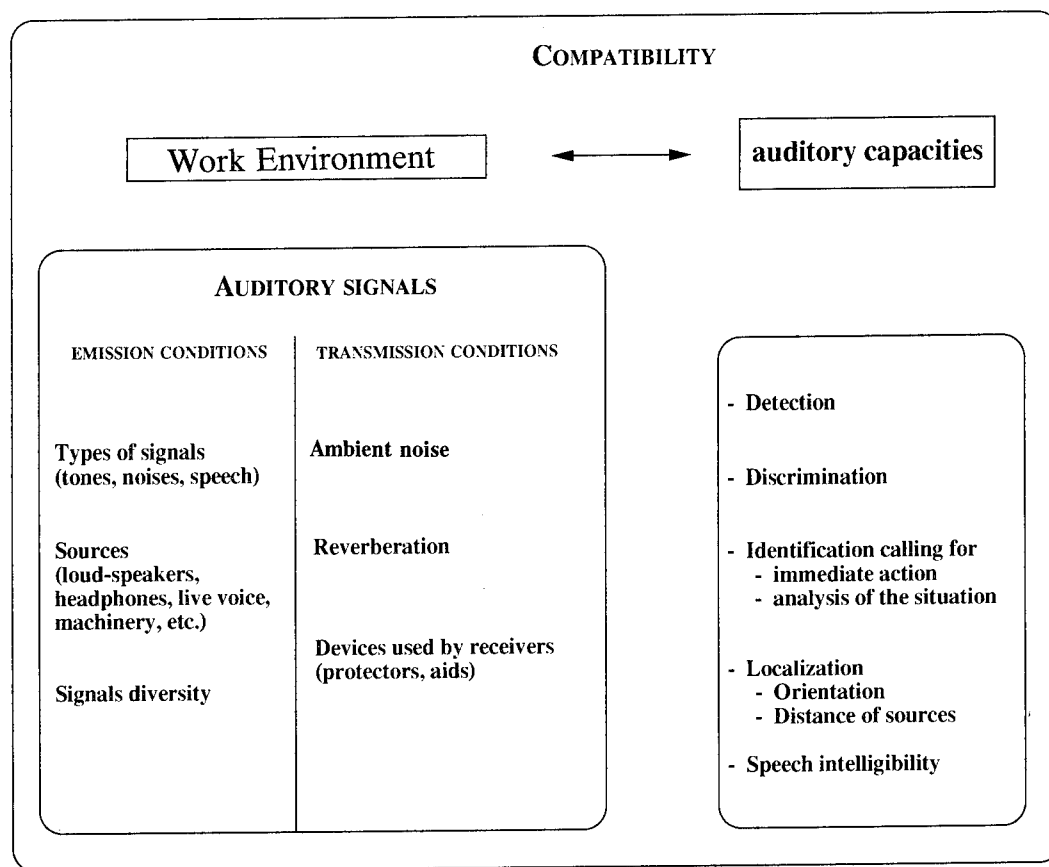


Figure 22-2 Proposed framework for systematic analysis of possible mismatches between auditory demands and capacities in the workplace.

tion, and localization of auditory warning sounds in such noisy and reverberant environments are very common demands.

Another type of demand is that of understanding live speech signals from coworkers and supervisors. Understanding amplified speech through paging systems or radio transmitters is also common. The speech signals are thus transformed by the limited bandpass and the distortion of the transmitter, and by the propagation conditions at the workstation, including the usual background noise.

Furthermore, very often workers are required to respond to such demands while wearing hearing protectors that can impose additional constraints on sound perception and verbal communication.¹⁴

When confronted with *normal* auditory capacities, most of these demands typically represent what ergonomists call “extreme con-

ditions,”¹⁵ namely, mismatches between demands and capacities. There are mismatches because a sound environment characterized by high noise level and long reverberation time restricts the possibility of making use of auditory temporal, spatial, and frequency resolution.¹¹ If the auditory capacities of workers with *normal* hearing are seriously challenged by the current environmental conditions in industry, *hearing-impaired* workers may be expected to be at a serious disadvantage.

Performance Impairment Among Workers With NIHL

The handicapping effects of NIHL in the workplace have yet to be investigated. This is paradoxical, considering the high prevalence of this occupational health problem in industry. But indirect evidence is available from

interviews conducted within the framework of studies on the consequences of NIHL⁵ and from answers to a handicap questionnaire¹⁶ by the 48 participants in experimental rehabilitation sessions.¹⁷ The workers came from a variety of industrial sectors and had hearing losses that met current criteria for compensation in Québec, namely a loss ≥ 30 dB on the average for 0.5, 1, 2, and 4 kHz. There were three items referring to work on the questionnaire. Problems related to task performance were reported by almost two out of five respondents. Half of the workers felt that social participation during breaks was felt as being restricted. Career advancement was an issue related to hearing impairment for more than one-third. These answers give clues about performance impairment in the workplace. In-depth analysis of the situation in different trades and industries should provide a more differentiated picture of the nature and severity of the difficulties experienced in the work setting. Meanwhile, performance impairment can be assumed when typical auditory demands in industry are confronted with hearing impairments associated with NIHL.

Signal Detection

Because of the loss of sensitivity alone, auditory signal detection may be impaired among workers with NIHL, especially if they wear

ear protectors.¹⁸ Let us consider the following example:

1. a background noise which has a third-octave band level of 80 dB sound pressure level (SPL) at and below 1 kHz with a downward slope of, say, -10 dB/octave above that frequency;
2. the use of ear plugs with an attenuation of 20–25 dB below 1 kHz and up to 35 dB around 3 kHz;
3. a worker who sustains a 60 dB loss of sensitivity at 4 kHz; and
4. the unmasked frequency components of the signal are above 1 kHz.

In such circumstances, the detection threshold for that signal will be determined by the combined absolute hearing sensitivity and protector attenuation at the target frequencies. In the illustration shown in Figure 22-3, the detection threshold is raised by more than 20 dB at 3.15 kHz due to protector use.

A somewhat similar situation could arise when a worker with a hearing impairment wears a *head* protector that can alter the acoustic features of incoming signals. A systematic investigation is underway in our laboratory concerning the effects of various types of hoods and other head protectors. Preliminary tests show, for example, that a welder's mask can act as a partial enclosure and create a condition of performance impairment, especially

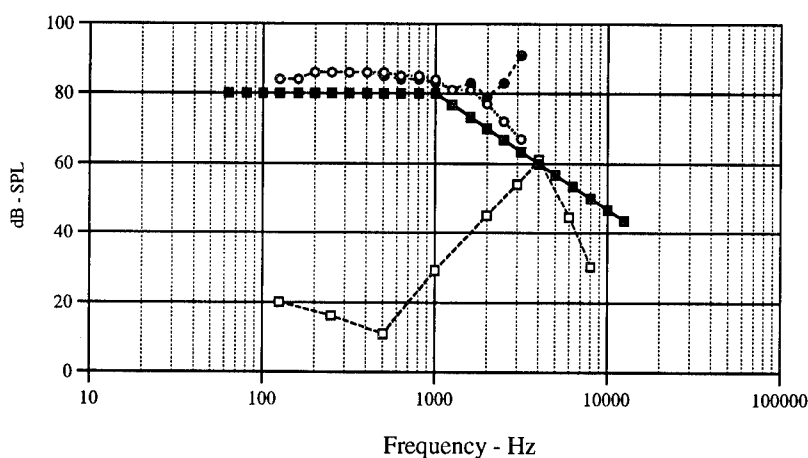


Figure 22-3 Example of signal detection by a hearing-impaired individual in a noise with a sloping spectrum: (□) Absolute thresholds; (■) noise spectrum; (○) unprotected detection thresholds in noise; (●) protected detection thresholds in noise.

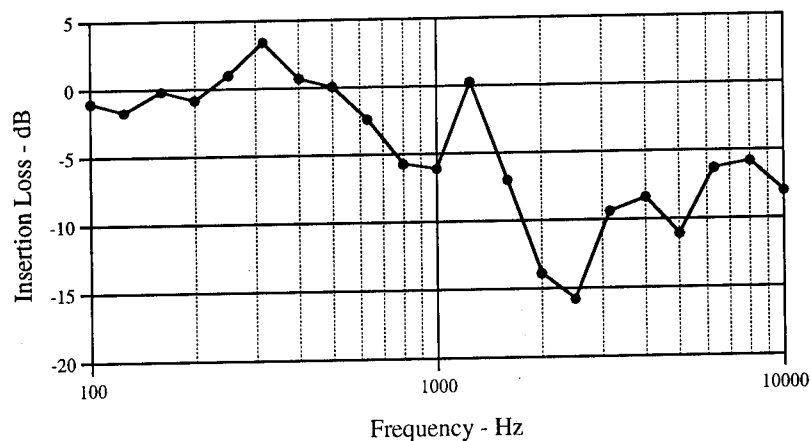


Figure 22-4 Insertion loss from a welder's mask at an angle of incidence of 0° in a freefield measured with a mechanoacoustic head simulator.

in workers with a hearing loss. This is illustrated in Figure 22-4. Figure 22-4 represents the insertion loss resulting from the use of a welder's mask, as measured on a mechanoacoustic head simulator.¹⁹ At specific frequencies, the mask can act as an effective barrier. As a result, with a noise spectrum and hearing loss configuration as in Figure 22-3, detection threshold may result from wearing the mask. As shown in Figure 22-5, this would be the case between 1.25 and 2.5 kHz.

A more common situation involves elevation of masked thresholds because of a loss of frequency selectivity.⁴ As shown below, this may require signal-to-noise ratios 15 dB higher than those needed by normal listeners

in the midfrequency range, and up to 25 dB at frequencies where marked asymmetry of auditory filters is found. The implication would then be that, with no systematic control over adjustment of the characteristics of auditory warning signals in industry, workers with NIHL would be more subject to accidents than normal hearing workers, because this may represent a serious handicap situation. Such a possibility is indirectly supported by evidence from a case-control study, conducted in a large shipyard, on the risk of accidents.²⁰ Three hundred manual workers who had had at least one injury were matched to 300 controls for age. Multivariate analysis for predicting injury identified job title, percentage of

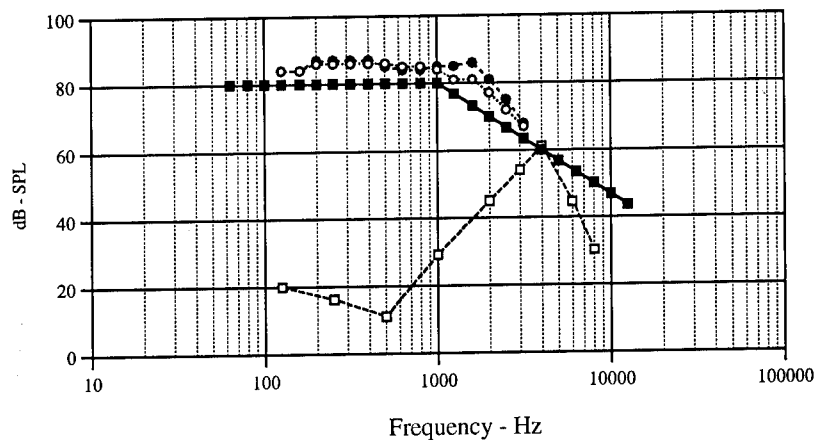


Figure 22-5 Example of signal detection with and without wearing a welders mask. Symbols as in Figure 22-3.

hours worked on the docks, exposure to continuous levels of noise higher than 82.5 dBA, alcohol consumption, and hearing thresholds at 4 kHz higher than 20 dB. From a systematic comparison between the two groups, poorer hearing sensitivity was one of the most significant risk factors.

Signal Discrimination

As mentioned above, there is as yet no systematic description of specific demands in the industrial work environment, and signal discrimination is no exception. There is, however, one condition that is commonly reported by workers and that has been subject to quantitative investigation,²¹ that is, the perception of incidental warnings that inform an operator of a defect or of the status of the process. Such incidental sounds in fact constitute a variation in the ongoing noise of the machinery operated. This means that the worker must be able to perceive a more or less subtle spectral change in the noise. Findings from one field study showed that, in order to be recognized, incidental sounds need to be presented at a higher level (+5 dB in that study) than that of typical warning sounds such as a horn.²¹ Elevated frequency discrimination thresholds are common among people with sensorineural hearing loss,²² although they are not necessarily correlated with sensitivity.²³ However, without a more precise description of the acoustic parameters of actual incidental sounds that must be heard in workplaces, it is difficult to evaluate the constraints imposed by NIHL on their perception.

Signal Recognition

Signal recognition by people with sensorineural loss has not been the object of systematic studies. Performance impairment may be inferred, considering the signal features that are generally relied upon in the design of distinguishable signals. These clues are the time pattern and the harmonic structure of auditory signals.²⁴ NIHL should not, by itself, impair identification of the temporal pattern of

signals that are properly designed with minimum interpulse intervals of 100 milliseconds, as recommended by Patterson.²⁴ However, because most industrial environments are reverberant, such interpulse intervals introduced in the signal design may be dramatically reduced by the propagation conditions prevailing in work rooms.²⁵ This is illustrated in Figure 22-6 with a pulsed signal comprising

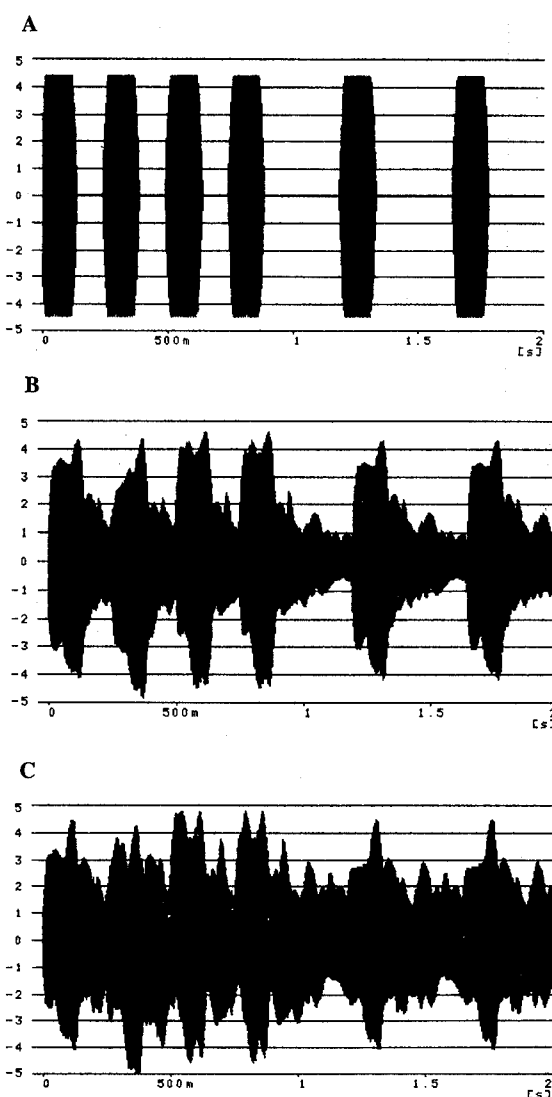


Figure 22-6 Pressure-time envelope of a pulsed signal. (a) signal as emitted at the source; (b and c) signal received in a simulated industrial room ($27 \times 27 \times 13$ m) with a reverberation time of 1 and 3 seconds, respectively.

four harmonic components with a fundamental frequency of 0.5 kHz.

The input signal was designed as proposed in the current guidelines.²⁴ It was made of series of six pulses, each one being characterized by a 25 millisecond rise and decay time with a total duration of 150 milliseconds. The first three interpulse intervals lasted 100 milliseconds, the last three, 300 milliseconds (Figure 22-6a). The signals comprised a total of eight harmonic components, the fundamental frequency being set at 0.5 kHz. A computer simulation of an industrial room measuring $27 \times 27 \times 13$ m high (total volume: 9473 m³) with reverberation times of 1 and 3 seconds was performed using the image method.²⁶ The impulse response at the center of the room was convolved with the auditory signal, whose source was placed at a distance of 1 m from one end of the rectangular space. The resulting time patterns are depicted in Figure 22-6b and c. Assuming a signal-to-noise ratio of 15 dB, the 100 millisecond interpulse interval is absent even with a reverberation time of 1 s (Figure 22-6b); the minimum sound level is 10 dB below the maximum level. Furthermore, the longer interpulse intervals are reduced to approximately 50 milliseconds in this condition. With a 3 second reverberation time (Figure 22-6c), interpulse intervals are not present anymore; the pulsed signal has been transformed into an amplitude modulated sound.

The above simulations indicate that interpulse intervals are systematically reduced in reverberant conditions. Knowing that temporal resolution is generally altered among hearing-impaired individuals,²⁷⁻³⁰ it can be expected that they will be more vulnerable to sound propagation effects on signal recognition. In particular, the amplitude of level variation for a gap to be perceived is markedly increased, up to 25 dB or more, when the interpulse interval is reduced.²⁸ This type of findings raises the need to investigate recognition ability in reverberant conditions among workers with NIHL when being presented pulse signals that are designed according to proposed ergonomic guidelines.

Localization of Sound Sources

Auditory signals are extensively used to warn of approaching vehicles such as loaders or lift trucks, and to warn of dangerous load displacements performed by cranes and hoists. Sound localization within noisy settings appears to be strongly determined by the frequency selectivity of the listeners.³¹ Loss of sensitivity in the relevant frequency bands reduces the perceptibility of high-frequency spatial cues, recently described in terms of directional transfer functions of the external auditory system.³² Noble et al.³³ found, in quiet listening conditions, an association (a) between vertical plane discrimination and high-frequency sensitivity and (b) between front-rear discrimination and mid- to high-frequency sensitivity. Distance perception in a noisy setting may be seriously impaired because of a compressed loudness function. Movement perception of sound sources that depends on interaural time difference, dynamic spectral, and level variations,³⁴ may also be seriously impaired because of NIHL. Clearly, there is a need for systematic investigation of the effects of NIHL on localization performance in noisy and reverberant environments that resemble industrial work settings.

Speech Perception

A considerable body of data has been collected on impaired speech perception in unfavorable listening conditions among people with sensorineural hearing loss.³⁵⁻³⁹ Performance in noise may be predicted using physical descriptors such as the Articulation Index^{40,41} and the Speech Transmission Index.⁴² The latter appears to provide a better prediction of the combined effects of noise and reverberation on speech intelligibility among hearing-impaired people. Because the industrial work environment is essentially characterized by the presence of relatively high levels of background noise and long reverberation times, it is almost a truism to state that hearing-impaired workers are faced with serious handicap situations whenever they are required to respond

to verbal instructions under such conditions. Further, they behave as if they are profoundly deaf when listening to speech through ear protectors in noisy environments.⁴³

What happens when such people are required to respond to a paging system, with a limited bandpass, that is operated in a highly noisy and reverberant industrial room? Similarly, how adequately can hearing-impaired workers respond to instructions given through a poor quality radio transmitter installed in the cabin of a noisy loader or truck, as is commonly the case in open-field mines or in locomotives? Such auditory demands need to be documented to ensure appropriate facilities that allow experienced workers with NIHL to pursue their careers without being confronted with handicap situations.

Cases of restricted advancement opportunities due to reduced communication skills were reported in a study presently being conducted with workers who had participated in experimental rehabilitation sessions¹⁷ 5 years earlier. Experienced workers with NIHL reported that they refrained from applying for promotion or were denied promotion because of their communication difficulties.

Matching Demands With Capacities

Auditory capacities are a given. This situation calls for efforts to adapt the work environment to the prevailing residual capacities of workers with NIHL. Noise reduction is of course the first step required to improve signal-to-noise ratios. Reverberation control is the second step. It not only provides more comfortable sound environments, but also reduces the level of constraint on temporal resolution and on speech intelligibility.

However, in many cases, these steps will not be sufficient to prevent performance impairment among workers with NIHL. Specific procedures for job accommodation are warranted when hearing loss is severe enough to prevent a worker from understanding speech transmitted via a paging system or a radio transmitter. Job accommodation is, in fact, an explicit provision of national legislations such

as the Americans with Disability Act in the USA,⁴⁴ the Canadian Human Rights Charter,⁴⁵ and others.⁴⁶ For those workers with NIHL who are currently employed, this means preventing or reducing the likelihood of their being confronted with handicap situations in realizing their careers.⁴⁷

As this perspective is relatively new in the field of NIHL, the need for development of practical tools that can help in adapting auditory demands in industry to individual capacities is considerable. One major area involves the development of a protocol for the design of auditory warning signals that can maximize detectability, recognition, and localization among workers with NIHL. Because detectability is a prerequisite for signal identification and localization, this needs to be given priority treatment. Consequently, we undertook the development of a procedure to define auditory signal features that are matched with individual capacities among hearing-impaired workers. The result of this endeavor is described below.

Predicting Individual Capacities for Auditory Signal Detection Amidst Noise Background

When a worker sustains serious NIHL, his/her frequency selectivity is almost inevitably affected. This means reduced ability to detect a warning sound amidst background noise, a typical requirement in the industrial work environment. To be able to adapt the working environment to the constraints imposed by NIHL, one must be able to characterize the residual capacity of signal detection in the presence of noise; in other words, given a certain ambient noise, one needs to predict the signal level at a given frequency that will meet the individual's detection capacity. This requirement calls for a clinical procedure that characterizes frequency selectivity in such a way as to generate predictions of masked thresholds, and a computer model that actually predicts individual detection performance. The study undertaken involved three

stages: first, adapting a laboratory procedure for measuring frequency selectivity to the constraint of a clinical test; second, adapting Detectsound[®] to individual rather than statistical predictions;^{18,48} and, third, validating such predictions among people with various degrees and configurations of sensorineural hearing loss.

Adaptation of Clinical Procedure

Our review of the procedures for measuring frequency selectivity⁴⁹ showed that auditory filter measurement with the notch noise procedure⁵⁰ was the most straightforward method of predicting masked thresholds for any signal frequency in the presence of any noise spectrum. The use of this method among workers with various degrees of hearing loss allowed us to adequately characterize auditory filters in most cases,⁴ with three types of areas of exception⁵¹:

1. indetermination of the filter width when its dynamic range is too small (i.e., <5 dB). In fact, there are cases of total loss of selectivity.⁵²
2. indetermination of the shape and the width of the filter when the loss of sensitivity is close to or above 60 dB hearing threshold level (HTL). This is due in part to the limit adopted for the masker level (i.e. 50 dB/Hz), in order to prevent annoyance and temporary threshold shift.
3. overdetermination of the shape of the filter in cases of pronounced upward spread of masking, leading to aberrant estimates of the filter slope on the high frequency side. These feasibility problems were addressed in our adaptation of the clinical procedure described below.

The Békésy method was adopted as the procedure for clinical measurement of masked thresholds. The standard error of measurement of masked thresholds among workers with NIHL was ≤ 2 dB.⁴

To limit testing time, the procedure was optimized, using only six masking noise conditions to characterize an auditory filter. As g_l

and g_u are the normalized lower and upper cutoff frequencies, the following masking conditions were adopted:

- absence of notch: $g_l = g_u = 0.0$;
- three symmetrical notch conditions:
 $g_l = g_u = 0.2, 0.3, 0.5$;
- two asymmetrical notch conditions:
 $g_l = 0.3, g_u = 0.5, g_l = 0.5, g_u = 0.3$.

The reliability of the filter parameter estimates with 6 versus 12 notch conditions was compared with measurements conducted with 16 normal listeners.⁵¹ The equivalent rectangular bandwidths (ERB) of the filters thus computed did not differ significantly, but the standard error of measurement with the simplified method was greater. It amounted to ± 7.5 Hz at 0.5 Hz, to ± 12 Hz at 1 kHz, and to ± 36 Hz at 3 kHz. This error corresponds to approximately 8% of the mean ERB. This was judged acceptable considering that only 20–25 minutes were required to characterize auditory filters between 0.25 and 4 kHz at one-octave intervals with the simplified procedure.

Normative data were collected with the simplified procedure at 0.25, 0.5, 1, 2, 3, and 4 kHz with the masking noise level set at 40 dB/Hz, and also at 1 and 3 kHz with the masking noise level set at 50 dB/Hz. Participants were 96 normal listeners (hearing threshold level⁵³ better than 15 dB between 0.25 and 6 kHz) between the ages of 18 and 26 years. Auditory filters were characterized using the rounded exponential model.⁵⁴ The resulting average and 95th percentile values of ERB are presented in Table 22-1. As expected, they tend to be systematically higher than those obtained under laboratory conditions. For instance, the average ERB at 2 kHz obtained in the laboratory⁵⁵ corresponds to approximately 80% of the value obtained with the simplified procedure.

For masked threshold prediction, this procedure involves a slight overestimation, that is, erring on the safe side, in the context of emergency signal detection. For judging the normality of auditory filters, the values in Table 22-1 are valid with ERB measures performed with proposed clinical procedure.

Table 22-1 Mean and 95th Percentile of ERB Values at Center Frequencies ($n = 96$)

	Filter Center Frequency (Hz)					
	250	500	1000	2000	3000	4000
Mean (Hz)	92	117	192	388	560	695
95th percentile (Hz)	123	137	219	465	670	821

Validation of Masked Threshold Predictions Among Individuals With Sensorineural Hearing Loss

Participants

Participants were sampled in order to represent different degrees of sensorineural hearing loss lower than 75 dB HTL and different audiometric configurations. They were recruited among the clients of a regional audiological rehabilitation center in Montreal, using the following *exclusion* criteria: air-bone gap > 10 dB between 0.25 and 4 kHz; abnormal tympanogram; interaural difference in hearing thresholds > 35 dB between 0.5 and 4 kHz; maximum loss between 0.25 and 4 kHz > 75 dB HTL⁵³; the over 65 or under 18 age brackets; presence of a disease associated with fluctuating hearing thresholds.

Inclusion criteria were determined according to audiometric configuration, that is, based on the difference between average hearing thresholds in the high frequencies (2, 3, and 4 kHz) and the low frequencies (0.25, 0.5, and 1 kHz). "Descending loss" was defined as a 10 dB difference between the high and low frequency average threshold, the inverse being the case for "ascending loss." "Flat loss" referred to a difference inferior to 10 dB. Four groups were thus recruited as follows:

1. 13 individuals with a descending maximum loss of between 35 and 55 dB HTL;
2. 21 individuals with a descending maximum loss of between 55 and 75 dB HTL;
3. 6 individuals with an ascending maximum loss of 55 dB HTL;

4. 12 individuals with a flat loss of between 30 and 55 dB HTL.

The descending loss group with a maximum threshold of 75 dB HTL was overrepresented, given that auditory filters would be measurable in about 50% of the cases at the frequency showing the higher threshold. Because ascending loss is rare, it was underrepresented in our sample.

Procedure

The experimental setup was identical to the one used in a previous study⁴ with white noise filtered by two low-pass and two high-pass filters connected in series. The resulting filtering slope was approximately +260 dB/octave on the high-frequency side and -260 dB/octave on the low-frequency side. The continuous notched noise was combined with a 250 millisecond pulsed pure tone repeated every 500 milliseconds that was generated by a clinical audiometer and presented to the subject by means of a TDH-50 earphone. Masked thresholds were assessed by Békésy tracking during the 30 s/notch noise condition.

Auditory filters were assessed at 0.5, 1, 2, 3, and 4 kHz in a random order. The masking noise conditions were also randomly presented except for the first condition, which was always the all-pass noise ($g_l = g_u = 0.0$). Testing was initiated with the masking noise level set at 40 dB/Hz. When the all-pass noise induced less than a 5 dB masking effect, the noise level was set at 50 dB/Hz. When less than a 5 dB masking effect was also observed

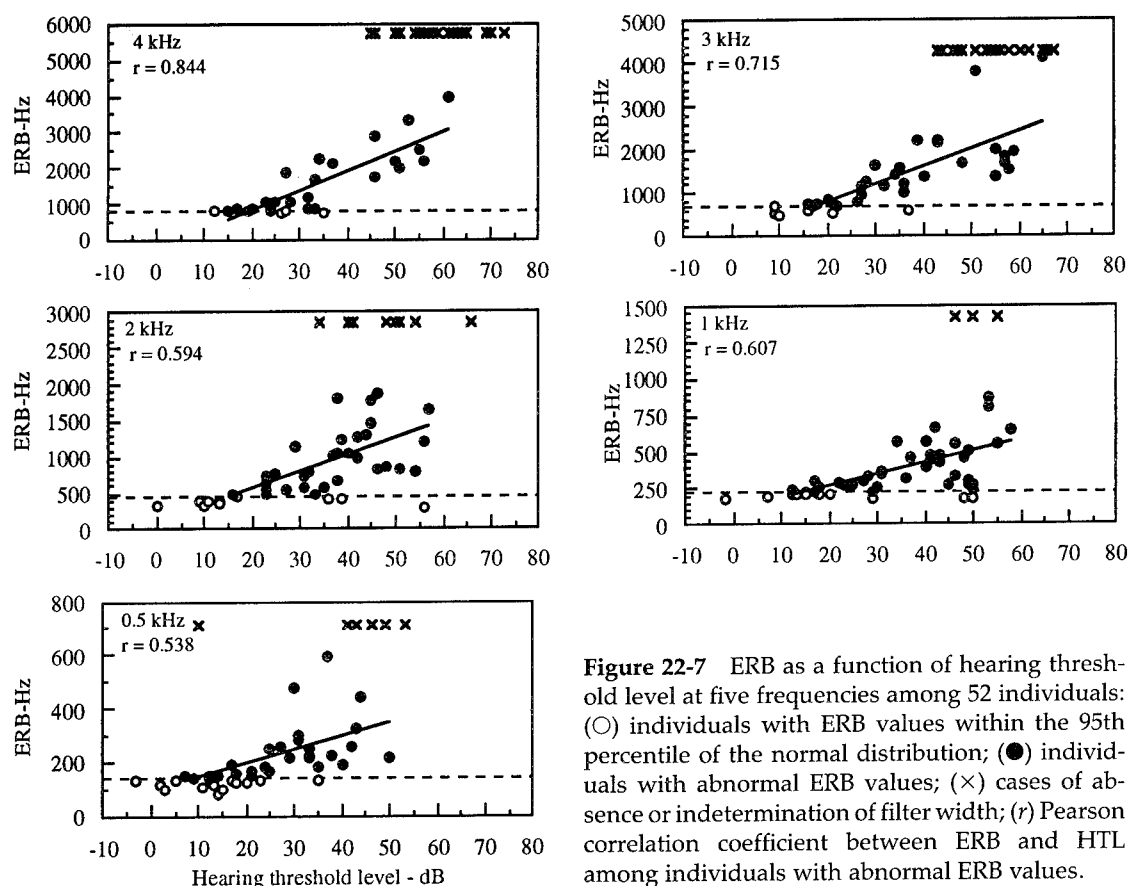


Figure 22-7 ERB as a function of hearing threshold level at five frequencies among 52 individuals: (○) individuals with ERB values within the 95th percentile of the normal distribution; (●) individuals with abnormal ERB values; (×) cases of absence or indetermination of filter width; (r) Pearson correlation coefficient between ERB and HTL among individuals with abnormal ERB values.

in the latter condition, the auditory filter width was considered to be “undetermined.” “Absence of auditory filter” was defined as a difference of less than 3 dB between masked thresholds obtained in the all-pass noise and in the widest notch condition ($g_l = g_u = 0.5$). To avoid aberrant estimates of the filter slope on the high-frequency side,⁵⁶ masked thresholds were also assessed with six high-pass noise conditions ($g_u = 0.0, 0.1, 0.2, 0.3, 0.4, 0.5$) when estimates of P_u resulting from notch noise testing were >50 .

The auditory filters were characterized using the mathematical expressions proposed by Glasberg and Moore.⁵⁴ When masked thresholds were obtained at 50 dB/Hz, the auditory filter characteristics were estimated for a 40 dB/Hz masking noise level based on the assumptions proposed by Glasberg and Moore, with variation in the filter slope restricted to the low-frequency side.

Characteristics of Auditory Filters

It was shown in previous studies^{4,57,58} that ERB tends to increase systematically with HTLs above a certain degree of hearing loss. Knowing the difficulty involved in precisely determining that degree of loss, it was decided to consider those cases with abnormal ERB values, that is, greater than the 95th percentile (Table 22-1) separately from those with normal ERB values. As shown in Figure 22-7, there is an obvious relationship between abnormal ERB and HTL. The linear correlation coefficient is statistically significant at the 0.01 level at the five frequencies tested. However, at a given HTL, large differences in filter widths are observed. For instance, at 2 kHz, a 45 dB threshold is associated with ERBs extending from 844 to 1868 Hz. These large individual differences, together with the pronounced widening of the auditory filter found

among individuals with limited hearing loss (i.e., <55 dB) support the use of *individual* prediction of masked detection thresholds with respect to acoustic signaling in the workplace. This is not the case, though, for severe hearing losses for which absence of frequency selectivity is observed or measurement of auditory filters is not possible within the limits inherent in the present procedure. As can be observed from Figure 22-7, the likelihood of being able to measure auditory filters decreases markedly above 60 dB HTL. As a matter of fact, filters could not be characterized when HTLs were ≥ 65 dB.

Another feature of the auditory filter data is its occasional pronounced asymmetry among individuals with hearing thresholds above 30 dB HTL. Figure 22-8 presents the individual values of filter slope ratios as a function of HTL. Cases of asymmetry with a more shallow high-frequency slope were associated with ascending audiometric configurations in the majority of cases. Conversely, instances of shallower low-frequency slopes corresponded, as expected, with descending audiometric configurations. Given that in such cases p_u values are derived from measurements of masked threshold with high-pass noises, such values represent a more valid estimate of asymmetry than with ascending audiometric configurations.

For the practical purpose of predicting masked thresholds, the proportionality constant K , introduced in the mathematical fitting of the auditory filter shape to the thresholds measured in different notch noise conditions,⁵⁰ becomes critical. This constant makes it possible to match the power of the masking noise that enters the auditory filter to the power of the signal at audibility threshold. It is assumed to be a measure of postfilter processing efficiency, which would not depend on filter characteristics. As illustrated in Figure 22-9, individual values of K show relatively large differences, extending, for instance, from -12.4 to $+12.6$ dB at 4 kHz.

The theoretical significance of this parameter has been seriously challenged, and alternative mathematical descriptions of auditory filters have been proposed.⁵⁹ But the latter

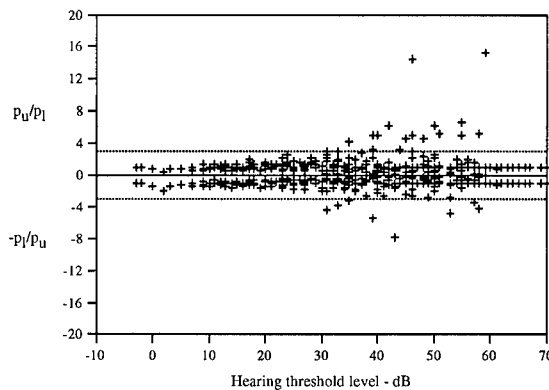


Figure 22-8 Distribution of individual p_u/p_l and $-p_l/p_u$ ratios as function of absolute hearing threshold ($n = 260$). The dotted lines refer to the 95th percentile among normal listeners.

description rests on fitting procedures that, in order to be reliable, require data from a large number of masking conditions. This is incompatible with our simplified testing procedure. For the purpose of predicting detection thresholds of auditory warning signals in industry, individual K values contributed significantly to minimize prediction errors as indicated below.

Based on the above description of auditory filters among people with sensorineural hearing loss, three parameters are expected to affect detection thresholds in noise: filter width, filter asymmetry, and postfilter processing efficiency. These combined sources of individual variation were expected to account for the variability in detection thresholds with different broadband noise spectra.

Detection Thresholds in Broadband Noises

To test the validity of predicted masked thresholds from individual auditory filter characteristics, the 52 participants were asked to detect pure-tone signals at 0.5, 1, 2, 3, and 4 kHz in three spectra of broadband noise at 85 dBA, in addition to the all-pass white noise at 78 dBA used in auditory filter characterization. The low-pass, high-pass, bandpass, and all-pass noise spectra depicted in Figure 22-10 were considered as valid replicas of prevailing ambient noise conditions in industry.

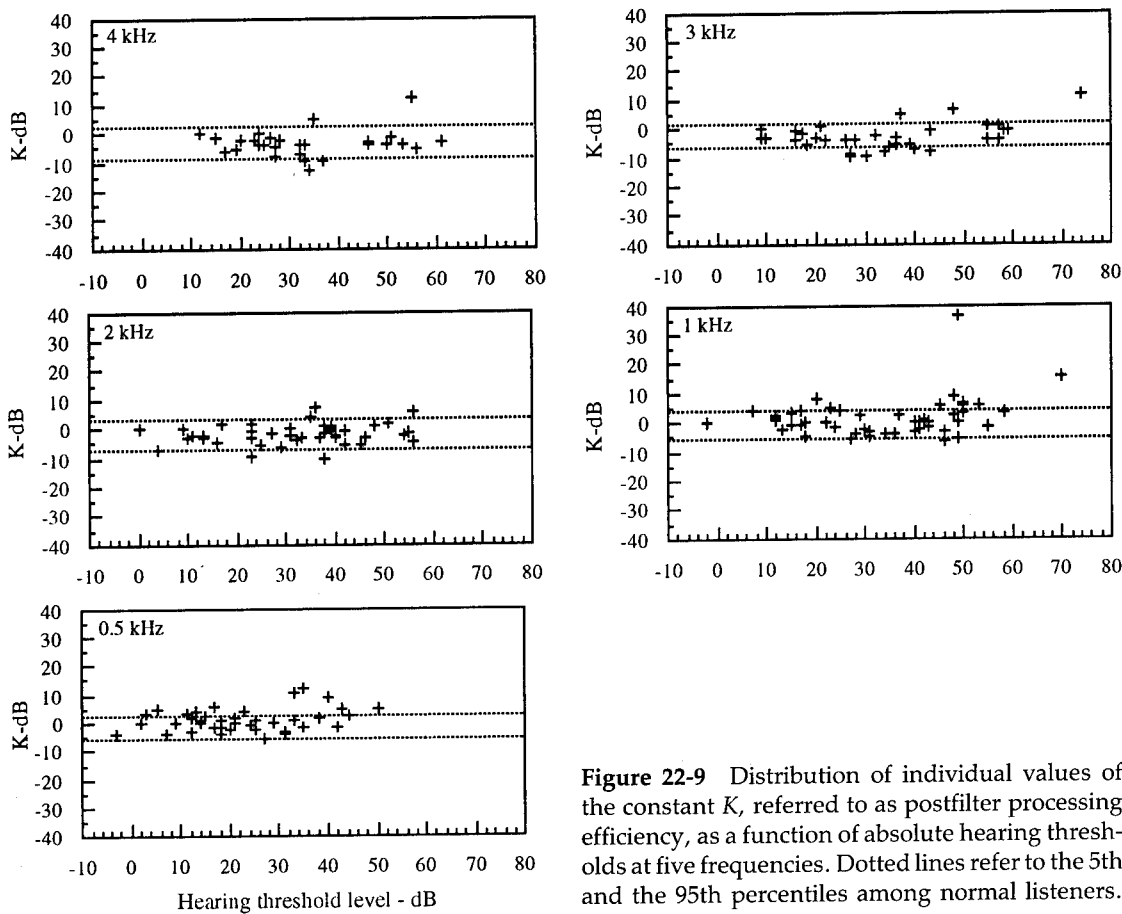


Figure 22-9 Distribution of individual values of the constant K , referred to as postfilter processing efficiency, as a function of absolute hearing thresholds at five frequencies. Dotted lines refer to the 5th and the 95th percentiles among normal listeners.

Considerable individual variations were observed in the detection thresholds with the four broadband noise conditions tested. This is illustrated in Figure 22-11 using the data obtained at 0.5, 1, 2, and 3 kHz, with the bandpass noise plotted against absolute hearing thresholds.

At 1 and 2 kHz, where the noise spectrum is flat, no relationship emerges between masked and absolute threshold level, as expected. There is nevertheless a 14 dB difference at 1 kHz and a 17 dB difference at 2 kHz between the higher and the lower masked threshold (except with one individual who deviates markedly from the rest of the group).

Where the noise spectrum is sloping, there appears to be an association between absolute and masked threshold, above a certain value of absolute threshold: namely, above 35–40 dB HTL at 0.5 kHz and above 50 dB at 3 kHz.

This is attributed to filter asymmetry and to upward spread of masking effects, which is more likely with poorer hearing sensitivity as shown above (Figure 22-8). Thus, 26 and 28 dB differences between extreme values are obtained at 0.5 and 3 kHz, respectively. With the low-pass noise showing a maximum slope around 3 kHz, the staggering of individual data is even more pronounced: a 41 dB difference is observed between extreme values. These large variations confirm the need for individual adjustment of auditory signals with respect to the residual capacities of hearing-impaired workers.

Validity of Individual Predictions of Masked Thresholds

Table 22-2 presents the mean and standard deviations of the differences between pre-

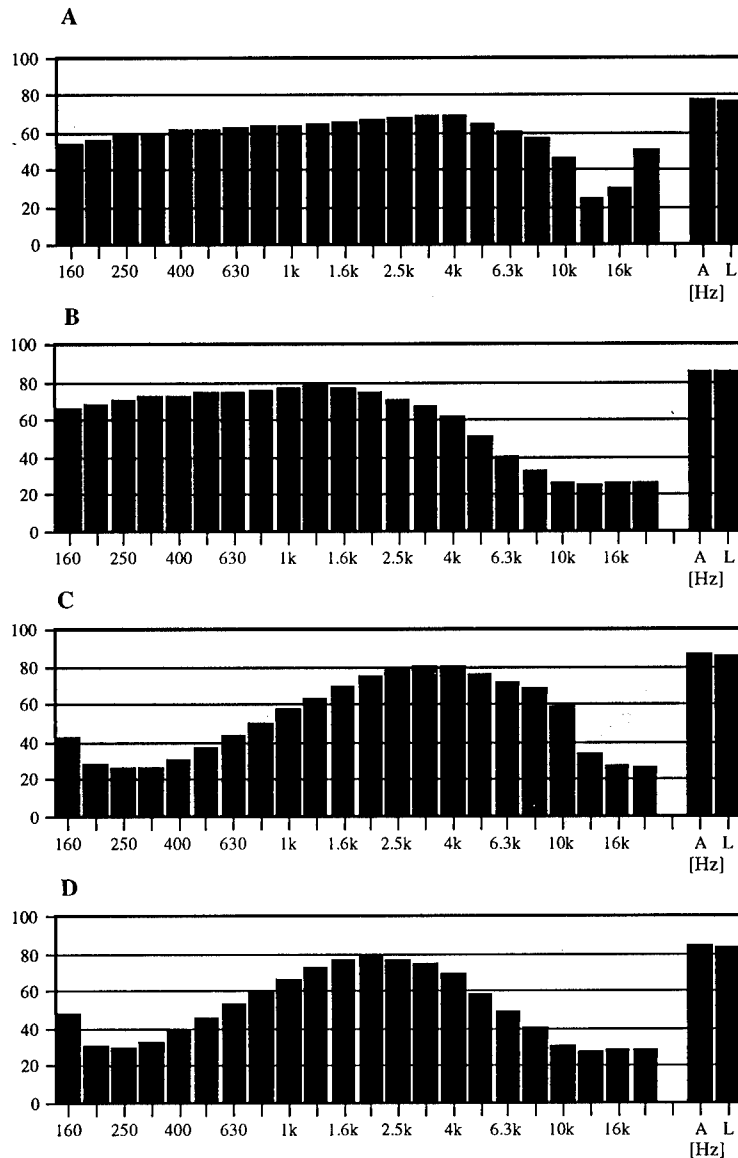


Figure 22-10 Third-octave band spectrum of the noises used to test the validity of masked threshold prediction among hearing-impaired listeners. The noises are referred to as (a) all-pass, (b) low-pass, (c) high-pass, and (d) bandpass. Overall unweighted (L) and A-weighted level is indicated on the right side of the graphs.

dicted and observed detection thresholds within the four masking-noise conditions described in Figure 22-10. Only those cases where the filter was actually characterized are included. As expected, the procedure generally led to slight overestimations of the masked thresholds, the average error of prediction being smaller than 2 dB, with one exception, that

is, 2.15 dB at 0.5 kHz with the white noise. The range in individual errors is relatively small (i.e., the standard deviations of differences were smaller than 4 dB), with two exceptions: at 4 kHz with low-pass noise, and at 2 kHz with bandpass noise. With all-pass noise, which actually served to assess the value of the fitting constant K , such errors are minimal.

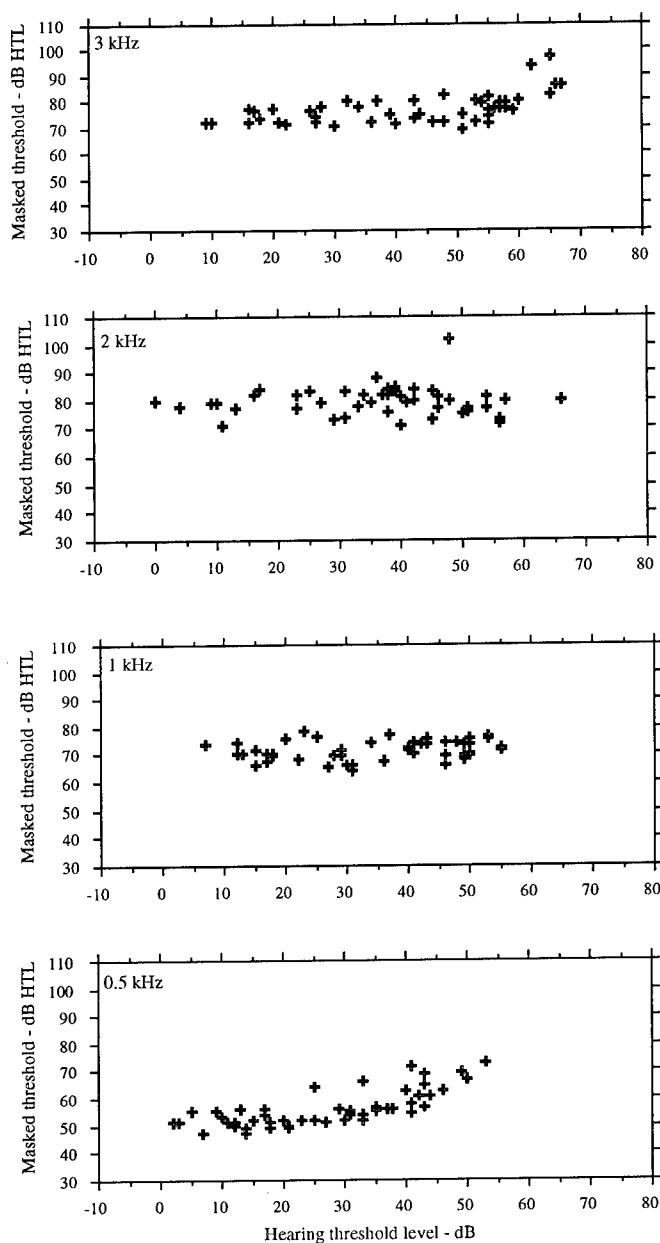


Figure 22-11 Distribution of individual thresholds masked by the bandpass noise at 4 pure-tone frequencies as a function of absolute hearing thresholds ($n = 52$).

As shown in Figure 22-12, predicted masked thresholds are systematically overestimated, with the error being more pronounced given lower absolute thresholds. This is attributable in part to the slight but systematic overestimation of ERB with the simplified auditory filter measurement procedure employed, which accounts for a 1.5 dB

overestimation of the masked threshold. The double exponential model⁵⁴ (ROEX p , r) used to describe the auditory filter also accounts for masked threshold overestimation, especially with individuals who have narrower filters. A more complex mathematical equation would be necessary to account for abrupt changes in the threshold of a pure tone in the presence of

Table 22-2 Mean and SD of Differences Between Predicted and Observed Detection Thresholds

	Signal Frequency (kHz)				
	0.5	1	2	3	4
<i>All-pass noise</i>					
Mean (dB)	2.15	1.78	0.93	1.36	0.63
SD (dB)	0.88	0.91	1.04	1.06	1.16
<i>n</i>	40	48	40	32	28
<i>Low-pass noise</i>					
Mean (dB)	1.08	1.31	0.53	0.76	0.66
SD (dB)	2.69	2.36	3.25	2.88	4.10
<i>n</i>	37	44	36	29	27
<i>High-pass noise</i>					
Mean (dB)	-0.01	-0.02	-0.44	1.66	0.47
SD (dB)	3.70	2.17	3.61	2.82	3.03
<i>n</i>	37	44	36	29	27
<i>Bandpass noise</i>					
Mean (dB)	0.41	0.86	-1.11	1.48	0.99
SD (dB)	3.89	2.61	4.41	2.70	3.91
<i>n</i>	37	43	36	29	27

n = number of individuals with whom the auditory filters had been measured.

narrow notch noises (e.g. $g_l = g_u = 0.2$) compared to threshold measurement in white noise ($g_l = g_u = 0.0$). For the practical purpose of adjusting the level of auditory warning signals, the prediction errors with this type of noise are negligible.

As can be seen from Figures 22-13 to 22-15, the magnitude of individual prediction error is typically 2.5 or 3 dB in a majority of cases, and is rarely above 5 dB, a value that is compatible with the audiometric measurement error. In fact, the proportion of cases of underestimation by 5 dB or more is equal to 0, 3.5, 4.6, and 6.4% for the all-pass, low-pass, high-pass, and bandpass noises, respectively.

The findings indicate that the errors tend to increase with signals located in the sloping portion of the noise spectrum. In those specific circumstances, the masked threshold depends not only on the overall width of the auditory filter, but also on its shape. Assessing the slopes of the filters involves much more uncertainty than assessing only their width. This is especially true when the filters tend to be highly asymmetrical, as is often the case with descending audiometric configurations.

Prediction of masked thresholds at 0.5 kHz in the presence of high-pass and bandpass noise tests the limits of our procedure because of minimal masking effects. While the third-octave band level measures 37 and 47 dB SPL at 0.5 kHz for high-pass and bandpass noise, respectively, it measures up to 80 dB where the level is at maximum. One can see in Figure 22-14 and 22-15 that the higher the absolute threshold and the wider the ensuing auditory filter, the greater the underestimation of predicted masked threshold at 0.5 kHz. As the auditory filters are narrower at lower sound levels, the excitation level in the inner ear probably depends less on the filter centered at the signal frequency than on those centered at higher frequencies, considering the slope in the noise spectrum. Estimation of the constant K at 0.5 kHz would consequently be misleading because of the influence of filters at higher frequencies. With abnormal filter widths, the constant K tends to be negative and this may lead to an underestimation of the masked threshold. The underestimation error is as high as 10 dB in some such cases. In actual industrial settings, such circumstances would

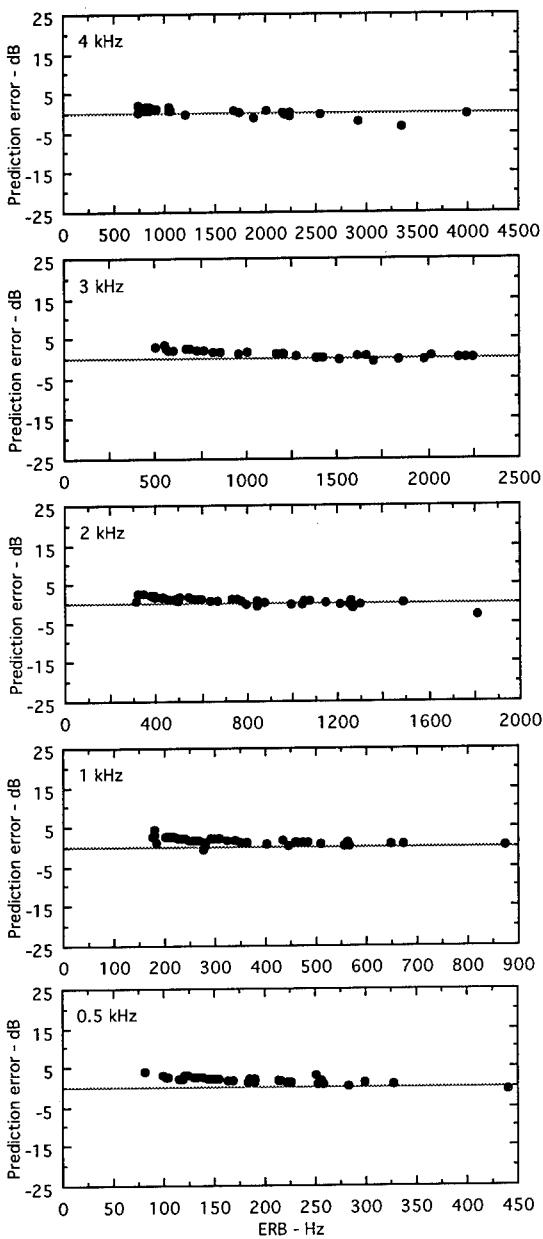


Figure 22-12 Individual values of the differences between predicted and observed thresholds masked by the all-pass noise at five pure-tone frequencies.

be extremely rare, however, because the slope of noise spectra is usually shallow,⁶⁰ and hearing loss is typically more pronounced in the higher frequencies.

Guidelines for auditory warning signal design prescribe signal level adjustment at 15 dB above the estimated masked threshold in or-

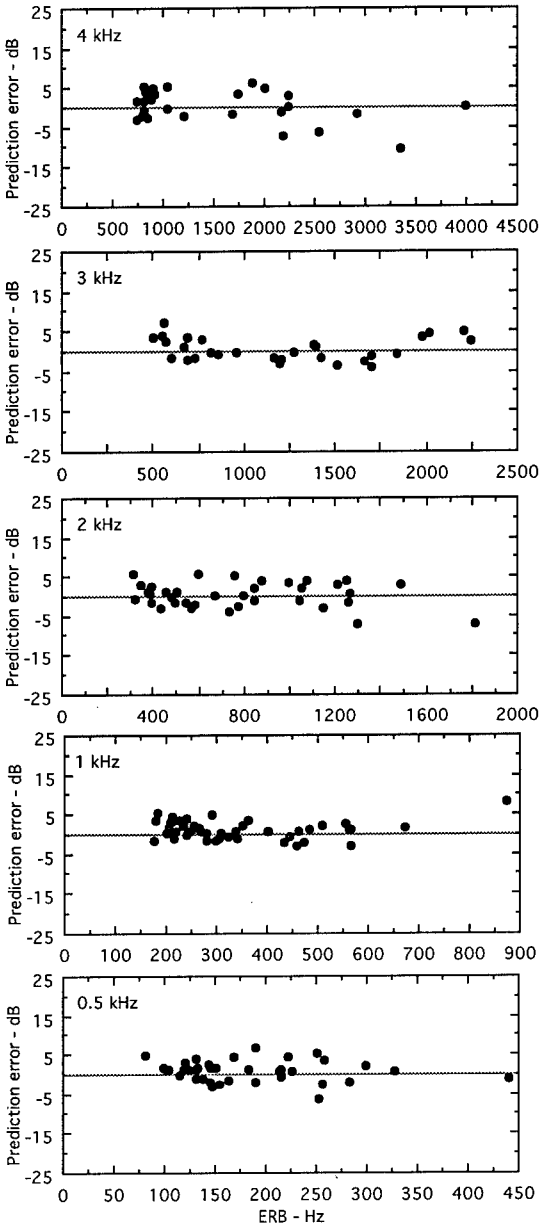


Figure 22-13 Same as Figure 22-12 with the low-pass noise as masker.

der to ascertain attention demand and facilitate signal recognition.²⁴ As errors of underestimation in individual masked threshold predictions are ≤ 5 dB for a very large majority of cases (Table 22-2), the use of the present procedure would ensure signal detection for almost anyone whose auditory filters have

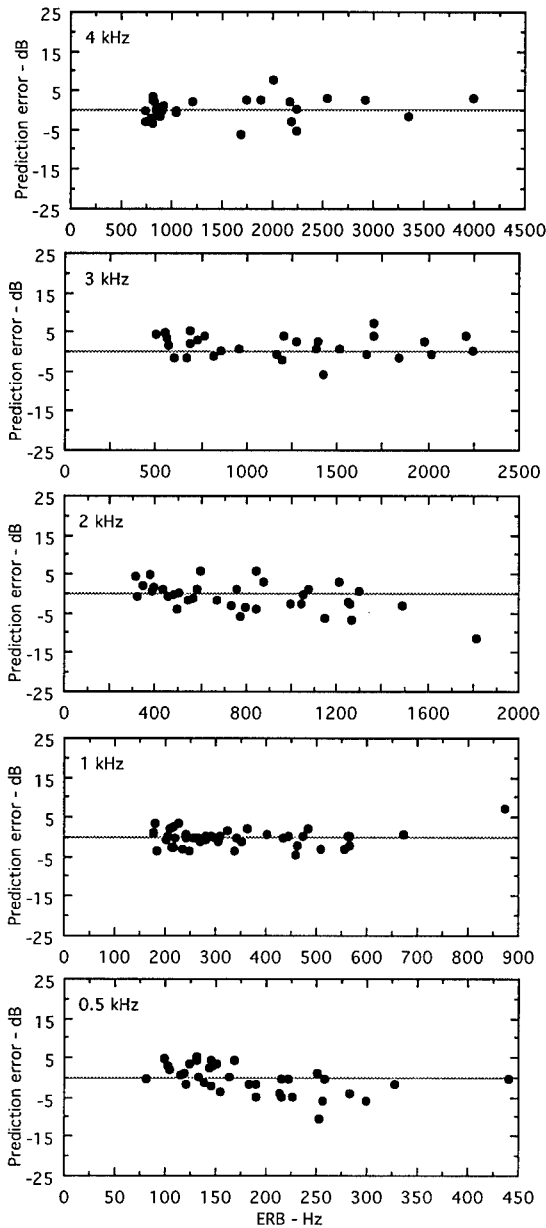


Figure 22-14 Same as Figure 22-12 with the high-pass noise as masker.

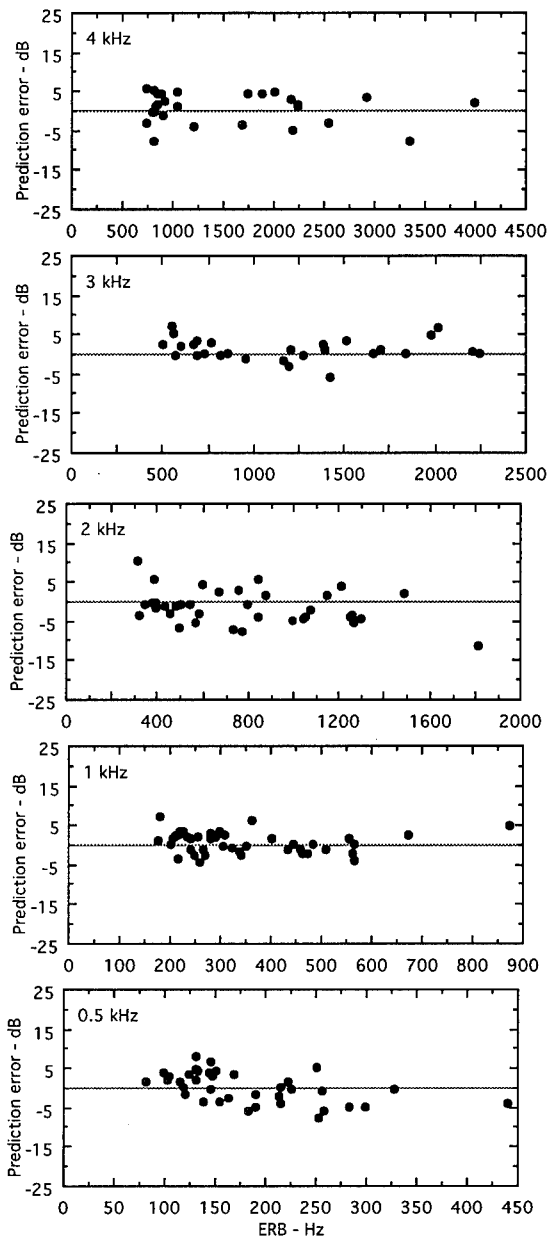


Figure 22-15 Same as Figure 22-12 with the band-pass noise as masker.

been characterized. Some individuals might, however, be at a slight disadvantage with respect to signal recognition.

Due to the constraints associated with the auditory filter assessment procedure adopted or the quasi-absence of filters at certain specific frequencies in some cases, there were

instances in which the filter width was simply estimated. This situation typically involved cases of hearing threshold levels ≥ 60 dB. In such cases, a value of 3 was assigned to p_1 and p_u and of -5 to the parameter r in such a way that the difference between the threshold measured in the white noise ($g_1 = g_u = 0.0$)

Table 22-3 Mean and SD of Differences Between Predicted and Observed Detection Thresholds

	Signal Frequency (kHz)				
	0.5	1	2	3	4
<i>All-pass noise</i>					
Mean (dB)	3.40	9.20	7.98	4.38	2.48
SD (dB)	5.38	—	4.96	6.03	3.21
<i>n</i>	10	3	8	18	23
<i>Low-pass noise</i>					
Mean (dB)	4.20	5.25	5.75	4.60	5.76
SD (dB)	6.00	—	5.15	9.10	6.46
<i>n</i>	10	4	12	19	21
<i>High-pass noise</i>					
Mean (dB)	-0.96	2.00	2.58	4.97	3.95
SD (dB)	4.59	—	8.35	6.34	5.09
<i>n</i>	11	4	12	19	21
<i>Bandpass noise</i>					
Mean (dB)	0.09	4.00	1.75	3.71	4.05
SD (dB)	7.27	—	7.62	7.38	6.46
<i>n</i>	11	4	12	19	21

n = number of individuals with whom the auditory filters had been estimated.

and the wider notch condition ($g_l = g_u = 0.5$) is equal to 3 dB. The ERB value is then equal to approximately 1.4 times the center frequency of the filter. Given that the constant K would become more negative with poorer hearing sensitivity and would lead to masked threshold underestimation, the constant was ignored in the predictions. Table 22-3 presents the mean differences between predicted and observed detection thresholds under the four masking noise conditions tested (Figure 22-10).

As expected from the large filter widths assigned to such cases, the predicted detection thresholds are systematically overestimated, except at 0.5 kHz with the high-pass and bandpass noise. The standard deviations of the difference between predicted and observed thresholds were also fairly large (Table 22-3). Instances of threshold underestimation of more than 5 dB were, nevertheless, very rare, occurring in < 5% of all the predictions. For safety purposes, it is preferable to overestimate than to underestimate the level of auditory warning sounds. The findings presented in Table 22-3 show that, even when hearing threshold levels preclude the mea-

surement of auditory filters with the proposed procedure, it is possible to determine the sound level signals to maximize their detectability for hearing-impaired individuals.

In more general terms, the present endeavor demonstrates the feasibility of adapting the most common auditory demand in the industrial workplace, namely, sound warning signal detection, to the constraints imposed by NIHL. A clinical procedure allows one to characterize, in 15–25 minutes of testing time, the residual capacity for signal detection in noise in the better ear of an individual who suffers hearing loss. A computer model⁴⁸ (Detectsound™) provides the required specifications in terms of signal level adjustment. The testing procedure together with the model constitute, in our view, a practical tool for job accommodation with people who sustain a hearing loss.

Concluding Remarks

A considerable body of knowledge is available for describing alterations in the auditory function of people with NIHL. However, consider-

ation of auditory performance from the perspective of actual work settings, in terms of incompatibility between workplace auditory demands and residual capacities, raises the need for much further investigation. In particular, a well-refined characterization of actual auditory demands in different industries and of the handicapping effects of NIHL in such settings is seriously lacking.

Given the general features of the industrial work environment, it is possible to pinpoint certain issues that need more systematic investigation among people with NIHL. These include the effects of noise and reverberation on signal recognition and localization, and the additional effect of wearing protective devices, including head protectors.

Investigation in such areas should lead to the development of job accommodation procedures that will prevent performance impairment among workers with NIHL. Adapting auditory warning signal levels to individual residual capacities, as proposed in the present account, could serve as a starting point for practical field trials of job accommodation for hearing-impaired workers.

More generally, any effort to reduce the noise level and reverberation time in industrial settings will benefit not only workers with hearing loss, by significantly reducing the main auditory demands, but also those with normal hearing who are forced by the presence of excessive noise to behave, to some extent, as if they were hearing impaired.⁶¹

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Chapter 23

Combined Effects of Hearing Loss and Hearing Protection on Sound Localization: Implications for Worker Safety

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Sound attenuating ear plugs and ear muffs are commonly worn in the industrial setting to protect against the injurious effects of noise exposure. They are generally accepted, some would argue, with insufficient justification,¹ as a low-cost, easily implemented alternative to noise reduction at the source. Unfortunately, selection is often based solely on a promised high degree of sound attenuation, even though extensive research has demonstrated that the real-world attenuation achieved by the wearer often falls short of the manufacturers' specifications.²⁻⁴ Less consideration is given to such variables as gender, hearing status, fitting technique, wearing time, the auditory perception tasks that will be carried out with the protector, or the characteristics of the noise background. All of these factors will affect successful utilization. It has been demonstrated that women have smaller ear canals, on average, than men. As a result, they may achieve significantly less attenuation when they wear plugs that are available in one size, designed for men.^{5,6} In men and women, some plugs are more likely than others to be dislodged from the ear canal with movement of the head and jaw because of their shape and materials.⁷⁻⁹ Achieved attenuation may also be compromised by poor fitting technique, particularly by alteration of the device to improve comfort.¹⁰

The wearing of hearing protective devices (HPDs) does not appear to interfere with audi-

tory signal detection and speech perception in noise in normal-hearing subjects. In contrast, performance on these tasks will be compromised in individuals with preexisting hearing loss.^{11,12} The sound attenuation of the device combines with the already raised hearing thresholds to create a more severe hearing handicap. It has been shown that level-dependent muffs with limited amplification may benefit these individuals.¹³ Such active devices amplify sound intensities below 85 dBA by 5–10 dB, and attenuate injurious intensities to the same extent as conventional level-independent HPDs. On the strength of this evidence, level-dependent protectors would seem to be a more appropriate choice for hearing-impaired users, particularly for situations characterized by sporadic high-intensity impacts superimposed on an otherwise relatively quiet background.

The present study explored the effect of HPDs on sound localization. The experiment was motivated by the concern that the wearing of these devices might interfere with the ability to discriminate the direction of hazard, resulting in an increase in the number of accidents in the workplace. Unlike the auditory detection and speech perception tasks that have been studied to date, sound localization has been shown to be adversely affected by HPDs in normal-hearing individuals. Muffs are more disruptive than plugs, probably because they interfere with cues to front/back

directionality of the source provided by the pinna.¹⁴⁻¹⁷ Level-dependent muffs that provide limited dichotic amplification do not appear to be different from conventional devices, although those with limited diotic amplification are severely disruptive.¹⁸ Potential benefits and drawbacks for hearing-impaired listeners have not been explored. With the ears unoccluded, hearing-impaired listeners are able to accurately localize frontal stimuli. The degree of high-tone hearing loss will determine the extent to which front/back discrimination will be disrupted.¹⁹

Rationale and Experimental Design

Sound localization in the horizontal plane was compared in two groups of subjects, 24 with screened normal hearing and 23 with bilateral high-tone sensorineural hearing loss, all over the age of 40 years. Individuals were tested with the ears unoccluded (UN) and fitted binaurally with conventional level-independent E·A·R foam plugs (PL) and E·A·R 3000 ear muffs (MI), and level-dependent Bilsom 2392 muffs (MD). A feature of the Bilsom muff selected is limited dichotic amplification of 5–10 dB in the region of 800–4000 Hz for sound intensities below 85 dBA. In previous research, we have demonstrated that for hearing-impaired listeners, such level-dependent devices serve to lower midfrequency hearing thresholds and interfere with speech intelligibility to a lesser degree than conventional devices.¹³

Within each of the four protector conditions (UN, PL, MI, and MD), subjects were required to localize one-third octave noise bands, centered at 500 and 4000 Hz. Inclusion of both low- and high-frequency test sounds allowed for an independent assessment of the utilization of interaural time of arrival and intensity difference cues, respectively. The two 300 millisecond stimuli were each presented in quiet and in a continuous background of 65 dB sound pressure level (SPL) white noise, at an intensity of 80 dB SPL. Differential effects of masking noise for protected auditory percep-

tion were previously documented for normal and hearing-impaired listeners.¹¹

Method

Subjects

The normal group, aged 41–58 years, was recruited by means of advertisements posted in a number of University of Toronto teaching hospitals. In these individuals, hearing thresholds, screened in the laboratory using a modified Békésy tracking procedure,²⁰ were on average 14 (± 5) and 20 (± 9) dB SPL for pure tones of 500 and 4000 Hz, respectively. The hearing-impaired subjects, aged 42–73 years, were patients of the Otolaryngology Clinic at Mount Sinai Hospital, Toronto. All were diagnosed as having bilateral high-tone sensorineural hearing loss, based on history, physical examination, and audiometric findings. Their hearing thresholds at 500 and 4000 Hz were, on average, 23 (± 6) and 54 (± 8) dB SPL. For both groups, the maximum allowable interaural difference in threshold at each frequency was 10 dB to avoid a possible right/left bias in sound localization.

Apparatus

The apparatus was described previously.²¹ Subjects were tested individually in a semi-reverberant double-walled chamber (IAC Series 1200) that met the requirements for hearing protector testing specified in ANSI Standard S12.6-1984.²² The ability to localize sounds was assessed using a single array of six loudspeakers (Realistic Minimus 3.5) surrounding the subject at a distance of 1 m at azimuth angles of 60°, 90°, 150°, 210°, 270°, and 330° that is, 60° apart.²³ The stimulus was produced by a noise generator (Briel & Kjaer 1405), used in conjunction with one-third octave bandpass filter (Briel & Kjaer 1617). Stimulus intensity was specified by means of a programmable attenuator (Coulbourn S85-08) and a set of integrated stereo amplifiers (Realistic SA-150). The amplifiers were balanced so that levels emanating from the six speakers were within 2 dB of each other at the listener's

center head position. Stimulus duration and envelope shaping, as well as trial by trial loudspeaker selection, were controlled by a Coulbourn Instruments modular system.

When applicable, prerecorded continuous white noise was presented by means of a cassette deck (Yamaha KX W900U) over a set of three 3-way loudspeakers (Celestion DL10) arranged behind the localization speakers to create a homogeneous sound field.²⁰ The intensity of the noise was set using a manual range attenuator (Hewlett Packard 350D) and integrated stereo amplifier (Rotel RA-1412).

The audio system was accessed by means of a personal computer (AST Premium 286) through IEEE-488 and Lablinc interfaces, and digital I/O lines. Subjects responded to each stimulus presentation by means of a laptop response box comprising a circular array of six microswitch response keys configured like the speaker array (see Figure 23-1).

Procedure

One block of 120 forced-choice trials was presented for each of the 16 experimental conditions. Within a block, the stimulus was presented through each speaker 20 times, with the restriction that the six speakers were chosen in successive randomized sets of six to balance the effect of practice. The unoccluded condition was presented first, followed by the protected conditions. The order of the latter were counterbalanced across subjects.

A trial began with a 500 millisecond warning light on the response box, followed by a 500 millisecond delay, and then the 300 millisecond stimulus. The subject's head and body were not physically constrained. However, subjects were instructed to fixate a straight ahead visual target each time the warning light occurred and to keep the head and body steady during the stimulus presentation. A maximum of 5 seconds was allotted for the response. Both the response azimuth selected and latency of response were recorded. Latency (accurate to within 1 millisecond) was measured without the subjects' knowledge for uncompromised accuracy. No feedback was given about the correctness of the judg-

ment. However, practice trials preceded each condition.

Results

The overall percentage of correct judgments observed for the eight protector by frequency conditions in the quiet background are presented in Figure 23-2. The data for the 500 and 4000 Hz stimuli are shown in two separate panels. Within the panel, the outcomes for the normal-hearing and hearing-impaired subjects are compared for each of the four protector conditions, that include sound localization with the ears unoccluded (UN) and protected with the conventional plugs (PL), muffs (MI), and muffs with dichotic amplification (MD). These results were obtained by averaging the percentage of correct responses for the six azimuths within subject, and subsequently collapsing across subjects. The standard deviations associated with the values plotted ranged between 7 and 18% across conditions, and were not systematically larger for the hearing-impaired group. A nested analysis of variance (ANOVA) applied to the data indicated that for the 500 Hz stimulus, regardless of group, the percentage of correct judgments observed in the unoccluded condition was significantly higher than in the protected conditions. However, the three protectors did not differ. Overall, the scores for the hearing impaired were significantly lower than those for the normal group, by 8%.

In spite of their hearing loss, all the impaired subjects could localize the 4000 Hz stimulus with the ears unoccluded. On average, their scores were significantly lower than those for the normal group by 13%. In contrast, only one-half of the impaired group were able to hear the high-frequency stimulus, and thus perform the localization task with the plug. Two-thirds of the group could complete the task with the conventional muff, possibly because it provided relatively less high-frequency attenuation than the plug. All 23 subjects could localize with the amplifying Bilsom 2392 muff, and in that condition their performance was not significantly different than that of the normal group. For those

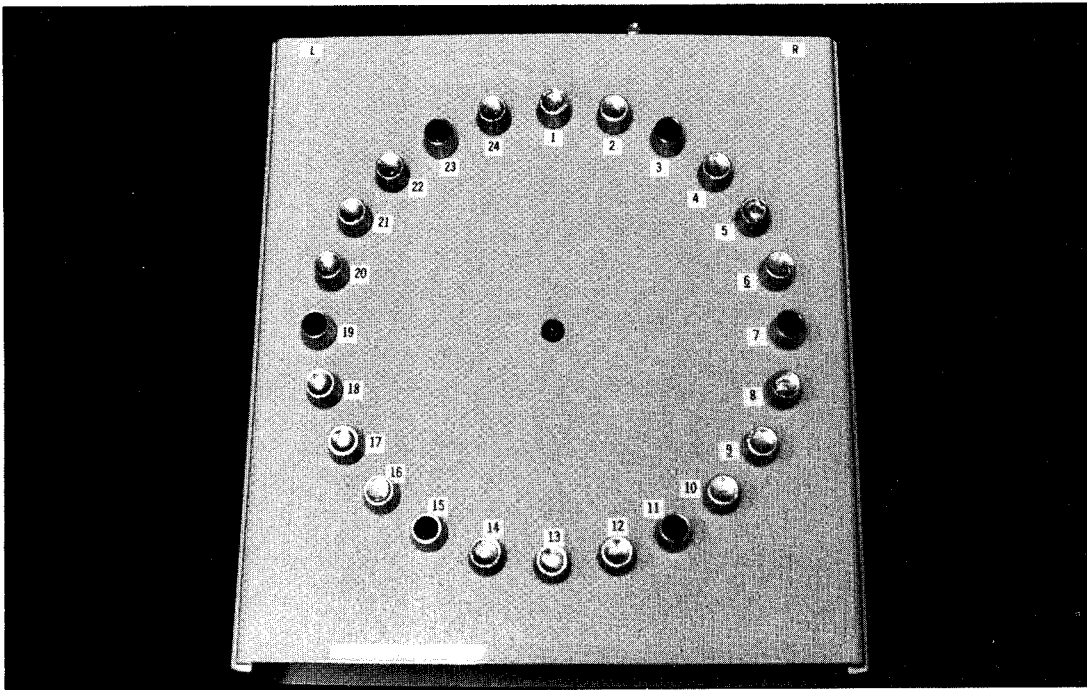


Figure 23-1 The response box used for sound localization.

hearing-impaired individuals who could not perform the task with any of the three protectors, a nonparametric paired comparisons statistical test showed that the outcomes did not differ. The normal-hearing subjects showed a decrement of 24% with either of the conventional HPDs relative to the unoccluded condition, and an additional decrement of 18% with the amplifying muff. In neither group did the

presence of the background noise have an effect on performance in the protected conditions for either stimulus frequency. However, in the unoccluded condition, a significant masking effect, amounting to a 10% decrement in score, was observed for the low frequency in the normal group. Details of the statistical analyses and outcomes are given in Abel and Hay.²⁴

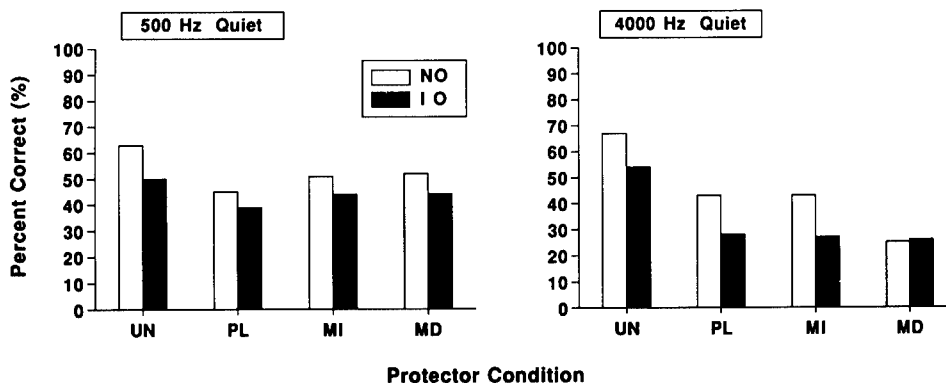


Figure 23-2 The effect of hearing loss on protected sound localization in quiet.

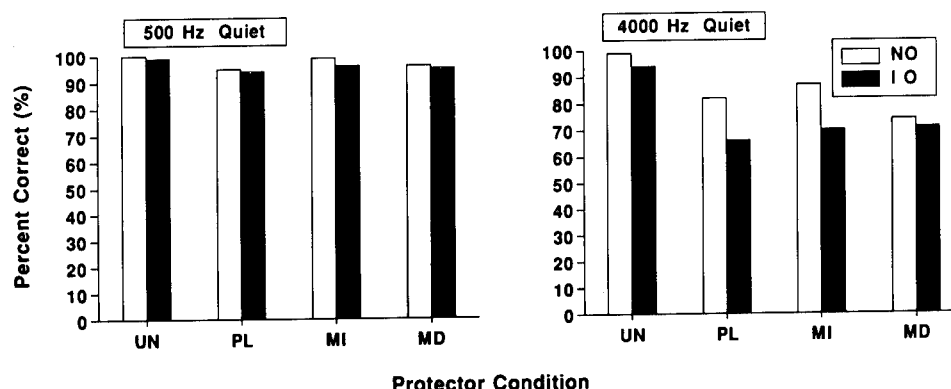


Figure 23-3 The perception of laterality: effects of hearing loss and hearing protectors.

To have a better understanding of how the three HPDs interfered with binaural cues in the perception of direction, the data were re-analyzed in terms of the correct discrimination of right from left and front from back. Figure 23-3 shows the percentage of trials in which the subject correctly judged the stimulus as coming from the right or left side, without regard to azimuth, for the eight quiet conditions. Both groups scored close to 100% correct for the 500 Hz stimulus. For the 4000 Hz stimulus, the pattern of results mirrored those obtained for percent correct averaged across azimuth, except that outcomes for each of the protected conditions were higher by 40–50%. A nested ANOVA on the results obtained by the two groups in the unoccluded and Bilsom 2392 muff conditions for the two stimulus frequencies, that is, conditions for which there were data available for all subjects, indicated that hearing loss was not a significant main effect. For the higher stimulus frequency, unoccluded performance significantly surpassed protected performance by about 24% in the two groups. With the amplifying device, right/left localization of 500 Hz was significantly better than localization of 4000 Hz by 23%.

Figure 23-4 shows the percentage of trials in which subjects correctly discriminated front from back, without regard to side. For this analysis, only the data obtained for the two front (30° and 330°) and two rear azimuths (150° and 210°) were included. A nested ANOVA applied to the data for the unoccluded and Bilsom muff conditions indicated

that hearing loss resulted in a significantly lower score of 13% when the ears were unoccluded, regardless of stimulus frequency. In spite of hearing loss, the impaired group, like the normal-hearing subjects, achieved significantly higher scores when discriminating front from back at 4000 versus 500 Hz. Unoccluded and protected performance were about the same at 500 Hz for the impaired group, 38%. At 4000 Hz, unoccluded performance surpassed protected performance for both groups. When protected, the impaired subjects were not different from the normal group. The mean protected scores for the four combinations of group and frequency were quite similar, ranging from 38 to 44%.

A question that arose in studying front/back discrimination was whether the wearing of the various devices might have induced a response bias or tendency to perceive all sounds as coming from a particular direction, regardless of the presentation azimuth. Because subjects had been informed that the test sound would emanate from each of the six speakers with equal likelihood, it was hypothesized that they would attempt to maximize the percent correct by minimizing response bias, that is, by utilizing each of the six response keys equally often.

Figure 23-5 shows the percentage of trials in which each of the six response azimuths was selected by the two groups, for the blocks of 120 trials presented for each frequency, by protector condition in quiet. If there was no bias, the percentage of responses associated

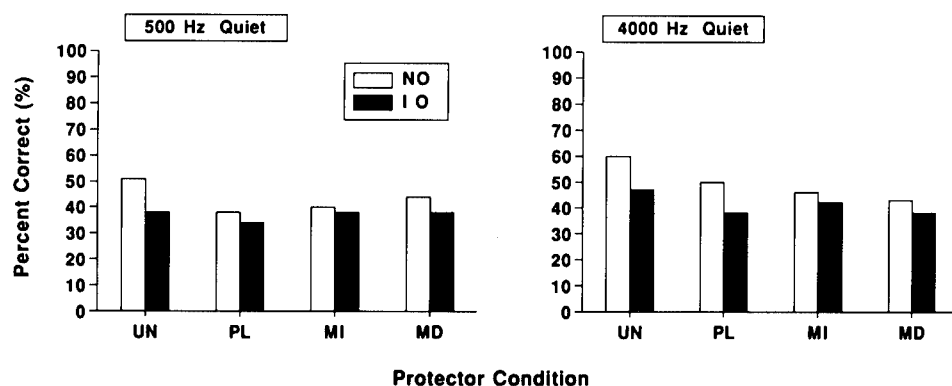


Figure 23-4 Front/back discrimination: effects of hearing loss and hearing protectors.

with each response azimuth would be 17%. Visual inspection shows that, in the case of the 500 Hz stimulus in the Bilsom muff condition, the normal group showed a slight bias of about 24% toward the two frontal azimuths at the expense of the two rear positions, which were each apportioned 9% of the responses.

For the 4000 Hz stimulus, both groups showed a relatively strong tendency to perceive the stimulus as coming from the back when they wore the Bilsom 2392 muff. The normal subjects showed a bias of 45% on average for the right rear azimuth (150°) compared with 25% for the left rear azimuth (210°). Fur-

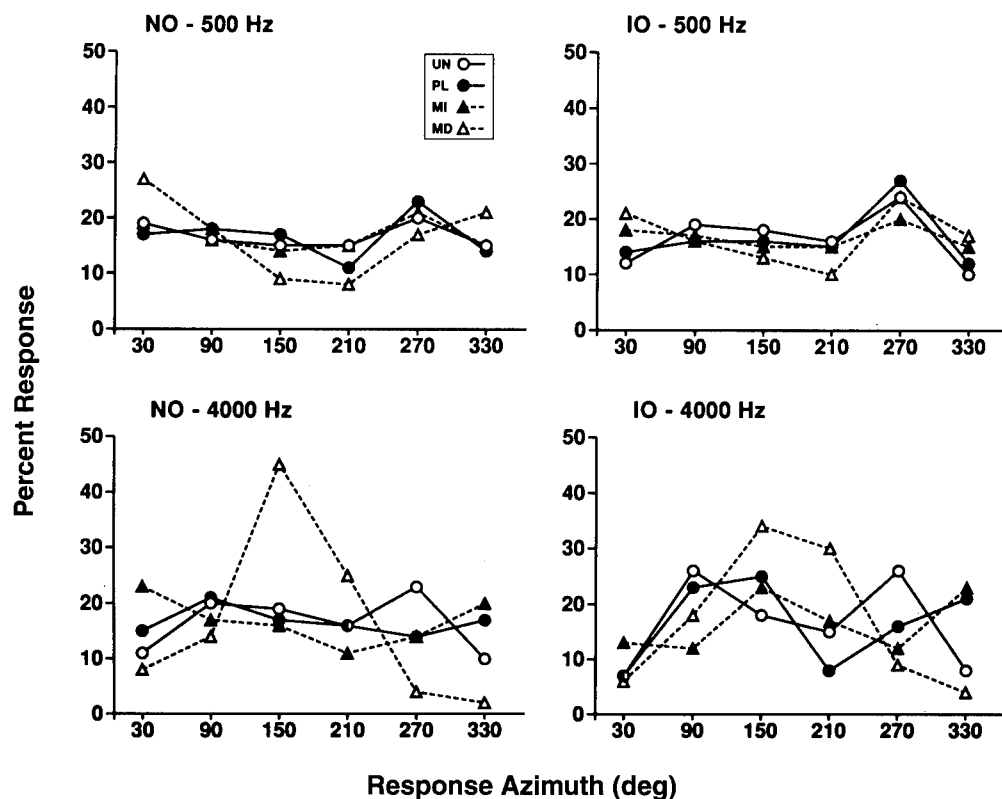


Figure 23-5 The wearing of hearing protectors and response bias.

Table 23-1 Sound Localization Response Times (ms) for Normal and Hearing-Impaired Subjects

HPD	Group	Response Type/Frequency			
		CRT		IRT	
		500	4000	500	4000
Unoccluded	NO	794 ± 203	846 ± 250	974 ± 353	1039 ± 401
	IO	871 ± 221	907 ± 236	1015 ± 307	1012 ± 338
E-A-R plug	NO	670 ± 202	677 ± 232	711 ± 231	746 ± 271
	IO	599 ± 273*	557 ± 303 [†]	620 ± 310*	588 ± 347 [†]
E-A-R 3000 muff	NO	640 ± 183	741 ± 223	689 ± 236	727 ± 242
	IO	688 ± 265	649 ± 380 [‡]	703 ± 314	674 ± 463 [‡]
Bilsom 2392 muff	NO	695 ± 304	750 ± 435	784 ± 410	812 ± 481
	IO	632 ± 250	538 ± 322	649 ± 280	575 ± 325

Values are mean ± SD. CRT, correct response time; IRT, incorrect response time. NO is the normal-hearing group, and IO is the impaired group.

* $n = 22$.

[†] $n = 10$.

[‡] $n = 16$.

ther analysis indicated that for at least one-half of the individuals in this group, there was right/left rear difference of 20% or more, favoring the right side. In contrast, the impaired group perceived the high-frequency stimulus as coming from the right or left rear positions equally often, on approximately 32% of the trials.

A measure that has not been previously examined in regard to performance with hearing protection is response latency.²⁵ In previous studies involving signal detection methodologies, we found support for the view that choice reaction time provides a way of assessing the nonsensory or decision-making component of perception.^{26,27} This response measure has been shown to be influenced by such variables as aging and task complexity.^{28,29} We reasoned that a study of the effect of HPDs and hearing loss on sound localization response latency might provide some additional insight into the way in which these variables might differentially affect sensory encoding and decision making.

The mean response latency for correct and incorrect trials for each HPD by frequency condition in quiet for each group is presented

in Table 23-1. These means are averages of median response times computed within subject. Medians are generally used in preference to means because of differences in the numbers of correct and incorrect trials across conditions, as well as skewness in the data.²⁶ As shown in the table, the average correct response time ranged from 538 to 907 ms. The mean incorrect response times ranged from 575 to 1039 ms.

As in our previous studies of auditory perception, we found that an incorrect decision took relatively longer to make than a correct decision. This difference, which was particularly evident in the unoccluded condition, may reflect an additional stage of cortical processing. ANOVAs indicated that for the correct response, hearing loss was not significant as an independent factor but was significant in interaction with the protector. Both groups took relatively less time to decide when wearing an HPD, but the type of device was not an important determinant of outcome. In the unoccluded condition, hearing loss resulted in a longer decision time. The reverse was generally the case in the protected conditions. Unoccluded versus protected listening, but not

hearing status, significantly affected the incorrect response times. The pattern of outcomes was similar to that observed for the correct responses.

Discussion

The results of this experiment on sound localization showed that both normal-hearing subjects and subjects with moderate bilateral sensorineural hearing loss were affected by the wearing of hearing protectors. In many respects, the pattern of outcomes was similar. None of the three HPDs tested had a negative impact on right versus left localization of a low-frequency stimulus, based on the interaural difference in time of arrival. Performance was close to 100% for both groups. In contrast, high-frequency lateralization based on the interaural intensity difference was compromised equally in normal subjects by the conventional muffs and plugs, but more so by the amplifying device. A high proportion of the impaired group were unable to hear the high-frequency stimulus with the conventional protectors, particularly the E·A·R plug that attenuated sounds to a greater degree than the conventional E·A·R muff. However, the two groups were equivalent when they wore the Bilsom 2392 amplifying device. In those impaired listeners who could complete the task with any of three HPDs, there was no difference due to device.

Front/back discrimination was also diminished by the wearing of the HPDs for both the low- and high-frequency stimuli. In the normal group, unprotected performance surpassed protected. However, there was no difference due to the device at 500 Hz. In contrast, at 4000 Hz there was an advantage for the unoccluded condition or when listening with the plug, compared with the two muffs. The protected localization sources at 500 Hz and the localization scores with the muffs at 4000 Hz were similar and close to 40% correct. This pattern of outcomes supports the view that when possible, subjects were utilizing the pinna cue for the higher frequency. The impaired subjects also showed an advantage for the higher frequency in the unoccluded condi-

tion, in spite of their hearing loss. Although their scores were on average lower than those for the normal group in the unoccluded condition, they performed at approximately the same level when they wore the amplifying device.

The results described above support the conclusions reached by Nobel et al.¹⁹ that both frontal and front/back sound localization may be maintained in hearing-impaired listeners, depending on the degree of loss. We found, on further analysis, that as long as the high-tone hearing loss was no greater than 50 dB SPL on average, the impaired subjects were able to perform the localization task when conventional hearing protectors were worn.²⁴ Although these listeners performed more poorly than normal under these conditions, their scores for the various protectors were no different, and thus they showed a relative benefit when wearing a muff that provided limited amplification of 5–10 dB.

The pattern of response bias was again similar for the two groups. An unexpected finding was the preponderance of rearward judgments with the amplifying device at the higher frequency. Normal subjects were more likely to perceive sounds as coming from the right compared with the left rear azimuth. One interpretation of this outcome is that the degree of amplification was not exactly the same for the two microphones in the pair of earcups. The absence of this asymmetry in the hearing impaired is not surprising. If there were an imbalance in the microphones, it would not be as readily detectable, given a moderate hearing loss. The 500 Hz stimulus was not in the frequency region of amplification, nor was it in the region of hearing loss in the impaired group. The normal group showed a tendency to perceive this stimulus as coming from the front. These induced bias patterns help to clarify the lower level of performance observed for the normal group for the amplifying compared with conventional HPDs.

The pattern of outcomes for response times was opposite to that for sensory processing. It was expected that when HPDs were worn, the task would be more difficult, and hence it would take longer to come to a decision about

the location of the sound sources. The reverse effect was observed. It is possible that the HPD resulted in a reduction in the number of perceptual alternatives. A decrease in the number of possible responses reduces response time.²⁸ As expected, the impaired subjects took longer to decide than normal, but only in the unoccluded condition. Their latencies were relatively shorter than normal in the protected conditions. This finding supports the conclusion that hearing handicapping variables in sound localization, whether from an actual hearing loss or induced by a protector, may result in a simplification of auditory space.

Taken together, the findings of this experiment support the conclusion that, unlike auditory signal detection and speech perception tasks, sound localization in normal-hearing subjects will be affected by the wearing of HPDs. This outcome is in line with previously reported results.¹⁴⁻¹⁸ Additionally, this study demonstrates that the various cues to localization in the horizontal plane will be differentially affected. Specifically, it appears that interaural intensity differences but not interaural time of arrival differences will be compromised. Muffs with limited dichotic amplification are comparatively more disruptive. As in earlier work, it was found that muffs resulted in greater interference with front/back discrimination than plugs, in spite of the greater attenuation of the plug. Our interpretation was that the pinna cue that contributes to high-frequency front/back judgments had been compromised.

Hearing-impaired listeners also had no difficulty in discriminating between right and left sources based on the interaural time of arrival cue when they wore protectors. This outcome may have been due to their relatively normal hearing in the low frequencies. As in the auditory tasks previously studied, these subjects had difficulty localizing the high-frequency stimulus when they wore conventional HPDs. In a high proportion, the combination of attenuation and moderate hearing loss rendered the stimulus inaudible. In our previous studies, we found that level-dependent protectors with limited amplification were less dev-

astating. This same relative advantage was apparent for sound localization. The high-frequency outcomes for the hearing-impaired and normal-hearing subjects were no different when the Bilsom 2392 device was worn.

The results support the view that the wearing of hearing protectors could be a disadvantage in the workplace if perception of the direction of hazard is an important consideration. In this circumstance, the data suggest that conventional earplugs would be the preferable choice for normal-hearing listeners because they are less likely to interfere with front/back discrimination. In contrast, hearing-impaired listeners would perform better with level-dependent muffs with limited dichotic amplification. However, this option has two important constraints. First, amplifying protectors would not be suitable for environments characterized by a continuous high-intensity noise background. In this situation, amplification might result in a greater risk of additional noise-induced hearing loss. Second, such devices appear to create the illusion of a rearward source, and this would impact negatively on front/back discrimination. However, because visual cues may accompany frontal sources, this drawback might be of less concern.

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Section IV

Human Studies of Noise-Induced Hearing Loss

Chapter 24

Extended High-Frequency Hearing Loss from Noise Exposure

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Extended High-Frequency Audiometry

Air conduction (AC) audiometry in the conventional frequency (CF) range is performed at octave intervals from 0.125 through 8 kHz. Threshold measurement above the latter frequency is termed extended high-frequency (EHF) audiometry. In EHF audiometry the sound source (transducer), the placement of the sound source relative to the ear canal, and the size and shape of the external ear/ear canal, are critical variables that have previously limited the clinical application of the method. These problems are largely due to the extreme directionality of high-frequency tones. Procedures with acceptable reliability were not established until some 30 years ago. Since then, different laboratories have largely employed different techniques, making it difficult to establish normative thresholds and recommended procedures. An ISO standard is still lacking for EHF audiometry (9–18 kHz), although this is now in preparation. Clinically, EHF audiometry is of value because of its extreme sensitivity in the early detection of cochlear pathology, because the pathological process tends to start in the basal, high-frequency region as a result of ototoxic, genetic, and age effects, and possibly noise-induced hearing loss (NIHL).

Early History

Fletcher¹ presented a detailed review on the early history of EHF audiometry. The upper range of hearing in humans and animals was

first studied in the 1870s by the British multi-subject genius Sir Francis Galton, the cousin of Charles Darwin. He invented a whistle, a narrow metal tube, the length of which could be varied by screwing a plug in or out, and claimed to produce frequencies between 6161 and 84 000 Hz. He admitted, however, that calibration was unreliable above 14 000 Hz.² A hydrogen-activated variant of the whistle was developed to reach even higher frequencies.³

In 1929 Fletcher⁴ studied the audible range of hearing, but concluded that the data were uncertain, with great individual variation above 10 kHz. Guild⁵ and Crowe et al.⁶ correlated high-frequency hearing loss with cochlear histology, testing mostly up to 8192 Hz, in some cases up to 16 384 Hz. Sivian and White⁷ tried to compensate for the tone directionality by comparing minimum audible pressure at the tympanic membrane, and minimum audible fields by monaural listening facing a free-field sound source, for frequencies 100–15 000 Hz. They reported problems with field determination for the higher frequencies. The audibility of high frequencies was also studied by Gavreau.⁸ Guild is credited with a lost study on high-frequency hearing in school children.⁹ In 1952 Dadson and King¹⁰ made one of the first serious attempts toward proper calibration, comparing the (physical) voltage at the earphone with the minimum sound pressure level (SPL) audible to the subject. With this reference they studied young, otologically normal adults at frequencies of 80–15 000 Hz and found that 90% responded within a 25 dB range below 4 kHz, within a

30 dB range at 6–12 kHz, and within a 35 dB range at 15 kHz (in the subjects able to hear 15 kHz), thus approaching acceptable reliability levels.

Early Normative Studies

In the 1960s Rudmose¹¹ developed a tapered plug which improved directional coupling of the transducer to the subject's ear canal, allowing reliable testing up to 18 kHz. He then determined the average hearing thresholds for otologically healthy young high-school students, and for each frequency used the voltage level recorded at each average threshold as reference for calibration. Fletcher,¹² having confirmed the improved reliability of this procedure, then studied CF and EHF hearing thresholds (up to 18 kHz) in otologically healthy, non-noise-exposed 6th, 9th, and 12th-grade school children.¹³ The data showed better EHF hearing in females than males and, better in the left than in the right ear. Corresponding studies were performed in 10–12-year-old healthy children,¹⁴ adults,^{15,16} and in normal-hearing 17–23-year-old males.¹⁷ Northern et al.¹⁸ then integrated these data with their own from non-noise-exposed adults, and in 1972 proposed the first recommended normative EHF thresholds.

Compared with these early normative data, patients surviving serious attacks of meningitis showed EHF hearing loss, even in cases with essentially normal hearing in the CF range.¹⁹ In patients taking ototoxic drugs, EHF hearing was found to be impaired earlier, and more profoundly, than below 8 kHz in humans²⁰ as well as experimentally in animals.^{21,22} The relationship between EHF thresholds and NIHL was studied by Sataloff et al.²³ and Downs et al.²⁴ The results of Corliss et al.²⁵ suggested that, in some subjects, NIHL could be first manifested in the EHF range.

Different Methods Applied in EHF Audiometry

In brief, the following EHF audiometry procedures have been employed:

- free-field loudspeaker systems,^{26–29}
- earphones,^{30–41}
- earphones with insert condenser microphone and Békésy fixed frequency tracking,^{42–48}
- various insert transducer systems,^{13,14,17,18,23,49–54}
- insert versus earphone systems,^{55,56}
- air versus bone conduction and masking,^{57–61} and
- electric bone conduction.^{62–67}

Different laboratories using different techniques makes it difficult to establish normative thresholds and recommended procedures. The lack of an ISO standard for EHF audiometry forces each laboratory to establish its own normative thresholds as reference for further study of pathological effects. ISO TC43/WG1 recently recommended an insert transducer system and restriction of the test frequencies to 10, 12, 13, 14, 15, and 16 kHz.⁶⁸

EHF Hearing: Nonnoise Effects

Test–retest reliability in the same subject corresponds to that in the CF range, whereas the intersubject variability is much higher.^{18,32,49,53,59} Age deterioration of hearing starts in the EHF range from the first decade of life, requiring different normative data for each decade.^{29,69} There are no EHF threshold differences related to sex, or to right/left ear thresholds.^{28,30,40,47} A recent review on ototoxic effects⁷⁰ states that most ototoxic agents initially produce basal cochlear degeneration before progressively involving lower frequencies. Thus, signs of ototoxic effects can first be documented by EHF audiometry. The high intra-individual test–retest reliability makes the method sensitive to deviations from baseline thresholds in serial audiometric monitoring of the individual patient. EHF audiometry is sensitive to damage induced by otitis media,^{33,71–75} otosurgery,^{76–79} and may predict the degree of otosclerotic stapes fixation.^{80,81} Tinnitus patients with normal hearing in the CF range show normal EHF thresholds,^{82,83} although Doménech et al.⁸⁴ reported elevated EHF thresholds in the majority of

tinnitus patients. Hereditary hearing loss may also be detected first in the EHF range.²⁹

Noise-Induced Hearing Loss

CF Range Effects

Exposure to sound/noise that exceeds a certain level leads to temporary (TTS) or permanent (PTS) hearing loss in the CF range, usually with a maximum around 3–6 kHz. Tone stimuli tend to produce a dip, or an additional dip, around 0.5–1 octave above the stimulus frequency.^{85–87} Repeated TTS > 25 dB may lead to PTS over time.⁸⁸ The threshold shift tends to increase with exposure time. The rate of CF NIHL development decreases as threshold shift increases.⁸⁹ Moderate noise levels characteristically produce asymptotic threshold shift (ATS): a more rapid initial loss rate, followed by a slower loss rate with time until the threshold shift stabilizes at the degree of hearing loss to be produced by that moderate noise level. ATS is demonstrated, for example in humans as experimental TTS from octave-band noise over hours,⁹⁰ as PTS from industrial noise over years,⁸⁹ and for impulse noise⁹¹; and in animals for both tones⁹² and impulse noise.⁹³ Compared with steady-state noise, impulse noise must be reduced by around 3 dB to produce the same degree of TTS.⁹⁴ Beyond a critical level, PTS may be produced by a single, short exposure, for example in humans for impulse noise,^{95,96} and in animals for tones⁸⁵ and impact noise.⁹³ Sound levels below 75 dBA are unlikely to produce CF range hearing loss even after long exposures.⁹⁷ Susceptibility varies across individuals.⁹⁸

EHF Range Effects

In a non-noise-exposed population of native Mabaans from the Sudan, EHF hearing (up to 24 kHz in 2 kHz steps) was comparable across age and sex, and thresholds were significantly lower than in Western age-matched controls, indicating that age effects on hearing loss recorded in the Western population may primarily be due to the cumulative effects of

noise.⁹⁹ In our society, age deterioration of hearing is detected in the EHF range from the first decade of life.^{29,54,69} Raised thresholds have been documented in noise-exposed high-school students,²⁵ in both children and adults,⁵⁴ and in adults.^{23,24,26} A number of the CF range effects referred to above are also demonstrated for the EHF range, but the variability across individuals appears higher in the latter.

Steady-State Noise

In paper-mill workers, the average threshold difference in the frequency range 10–14 kHz between noise-exposed and non-noise-exposed employees was around 19 dB for each frequency across age groups and exposure duration,²³ thus demonstrating asymptotic EHF threshold shifts for moderate noise levels. Moderate-level steady-state noise exposure tends to give threshold shifts first for the highest EHF frequencies, 13–20 kHz,^{100–102} and below 90 dBA changes may be restricted to the EHF range.¹⁰³ Across individuals with CF threshold shifts, EHF shifts may show considerable interindividual variation and may even be absent.²⁸ Paper-mill workers exposed to 80 phons showed asymptotic CF shift, but no EHF shift.¹⁰⁴ For disco and walkman music noise exposure, EHF threshold shifts may exceed CF shifts.⁷³

Half-octave or octave-related (harmonic?) basilar membrane phenomena have been recorded. Fritze and Köhler¹⁰⁵ have shown that low-tone 0.25 and 0.5 kHz exposure elicited four distinct dips in the TTS hearing thresholds: one octave above the exposure frequency, and around 4, 10–11, and 14–15 kHz, the highest frequency dip showing the longest decay time. This would indicate increased risk for PTS in the very highest range, which has subsequently been reported.¹⁰⁶ Noise-exposed workers with a 4 kHz PTS dip of 35 dB or less, showed related effects with two or more additional dips above 8 kHz and minimal shifts at other frequencies; for increasing 4 kHz dips, EHF hearing deteriorated, gradually concealing the EHF dips.¹⁰⁷

Impulse Noise

Acute acoustic trauma from high-level impulse noise characteristically produces a CF threshold shift at 3–6 kHz and considerable, even total, hearing loss in the EHF range with great interindividual variability.^{32,101–103,108–111} This has also been found histologically in the Rhesus monkey.¹¹² For more moderate exposure, threshold shift may be restricted to the EHF range from 15 kHz and above.¹⁰⁶ Beyond a certain critical noise level extensive high-frequency cochlear damage may be produced in humans from 2 to 3 kHz and up,^{32,102,106,111} and in animals for pure tones⁸⁵ and impulse noise¹¹³ (and R.P. Hamernik, unpublished data, 1994).

The Present Study

This study focuses on the relation between CF and EHF thresholds for subjects in different age groups with different CF grades of threshold shifts from steady-state and/or impulse (weapon) noise exposure. Due to lack of standardized normative data, EHF thresholds are given in dB SPL and compared with previously established normative dB SPL values from the same laboratory.⁴⁰ Median group thresholds of the noise-exposed subjects are given in Hallmo et al.⁴¹ The following presents in greater detail how EHF thresholds vary with progressive grades of CF NIHL. Characteristic features are illustrated by selected individual audiograms, and discussed in relation to asymptotic threshold shift (ATS), aspects of human and animal data, basilar membrane mechanics, and pathophysiological mechanisms at the hair-cell level.

Material and Methods

The study group consisted of 167 males, age range 18–59 years, median 45 years; in age groups 18–24 years ($n = 32$), 30–39 years ($n = 26$), 40–49 years ($n = 50$), and 50–59 years ($n = 59$). All had a history of occupational noise exposure, and a documented NIHL of at least 20 dB at 3, 4, or 6 kHz. For comparison, individual audiograms from occupational noise-exposed subjects aged 18–24 years with a 3–6

kHz CF hearing loss ≤ 20 dB, were also included among the case study audiograms. The majority of subjects were recruited among military personnel, conscripts and professionals, or were civilians. All were exposed to steady-state and/or impulse noise, with repetitive exposure to gunfire, jet aircraft runway noise, or machinery. Acute acoustic trauma from single impulse noise exposure was excluded from the study group. Other causes of hearing loss such as heredity, previous ear disease, ototoxicity, and head injury were excluded anamnesticly. Steady-state versus impulse noise effects on NIHL turned out to be hard to specify, as preexposure audiograms were largely lacking, and most subjects had been exposed to weapon noise during compulsory military service. Pneumatic otomicroscopy was unremarkable in all ears, and none presented an air–bone gap in CF. Interviews revealed that use of ear protectors had been minimal in the older groups in the earliest years of employment, and had not always been used, although compulsory, in occupational noise exposure during the last 10 year period.

AC and bone-conduction (BC) thresholds were determined in the CF range, and supplemented by AC thresholds in the EHF range, in 1 kHz steps, using the Interacoustic AS10HF tone generator, Koss HV/1A headset, and 1/3-octave band contralateral masking by a Brüel and Kjær random noise generator.⁶¹ The CF hearing losses were classified as Grades I–IV according to Man et al.¹¹⁴ Median CF thresholds (in dB hearing level, HL) and EHF thresholds (in dB SPL) were calculated for each age group and grade of CF NIHL.⁴¹ Different individual features of NIHL EHF threshold development and variability relative to progressive grades of CF NIHL were then reexamined and compared with normative thresholds.⁴⁰

Results and Discussion

Median group data⁴¹ showed that for each grade of CF NIHL, the EHF thresholds throughout were higher than in age-matched subjects not exposed to noise. Within age groups, NIHL grades and EHF shifts did not

correlate significantly with longer noise exposure time, indicating that asymptotic threshold shift (ATS) also exists in the EHF range. For slight to moderate CF acoustic trauma, NIHL Grade I, the EHF threshold elevations increased with increasing age. For more severe CF acoustic trauma, NIHL Grades III and IV, the EHF thresholds were poor and largely overlapping across age groups.

A more detailed analysis of these group thresholds revealed that for slight to moderate acoustic trauma, CF NIHL Grade I, the median EHF curves showed accentuated shifts at 9 kHz for the youngest age group 18–24 years, and at 9 kHz and from 13 kHz and up for the age group 30–39 years. For NIHL Grade II, the youngest 18–24 year group showed a widened 9–11 kHz dip, and more accentuated loss above 13–15 kHz, whereas the curves were poor and overlapping from age 30 years. For more severe CF acoustic trauma, NIHL Grades III and IV, the EHF thresholds were poor and largely overlapping across age groups. In the age group 50–59 years, the EHF curve was elevated for NIHL Grade I compared to the normative threshold, but least for the highest frequencies, and the EHF curves were elevated and largely overlapping for CF NIHL Grades II–IV.

Individual audiograms confirmed these tendencies, showing corresponding EHF dips with accentuated shifts in the 9–10 kHz region and above 13–15 kHz (Figures 24-1 a–j). For gradually increasing CF NIHL, young individuals 18–24 years demonstrated progressive development of EHF hearing loss, including no CF plus slight EHF loss (figure 24-1a); slight CF plus slight EHF loss (figure 24-1b); asymmetric Grade I CF plus more symmetric EHF thresholds (Figure 24-1c and d); symmetric Grade II CF plus less EHF loss (Figure 24-1e); or asymmetric Grade I CF with asymmetric EHF loss (Figure 24-1f). Asymmetry could occur throughout the CF and EHF range, as in Figure 24-1g, one ear showing Grade I CF with slight EHF loss, the other ear Grade III CF and extensive ipsilateral loss from 2 kHz in CF through EHF. Some older subjects showed rather symmetric Grades I+II CF loss with symmetric EHF loss > 13 kHz (Figure 24-1h),

Grades I+III CF loss plus symmetric severe EHF loss (Figure 24-1i), and Grades II+III CF loss with, respectively, severe and almost total EHF loss (Figure 24-1j).

Case studies thus showed considerable variability across individuals, which may be due either to differences in noise exposure and/or individual susceptibility. Phenomenologically, EHF loss may be present without CF loss, or be lacking, comparable to, or exceed the CF loss.

In young subjects, NIHL may seem to produce an EHF hearing loss which increases progressively for increasing grades of CF loss until an EHF ceiling effect is reached, approximately at a CF Grade III loss, when the EHF curves almost overlap for all age groups. This shows a reduced rate of EHF threshold shift with increasing grades of NIHL, as found in the CF range.⁸⁹ The subject group 30–39 years reaches this EHF ceiling effect already at CF Grade II loss, and the older groups, 40–49 and 50–59 years, are close to the EHF ceiling already at CF loss Grade I, indicating that in the EHF range, NIHL susceptibility increases with age more than in the CF range.

For younger subjects, NIHL may be first manifested as an EHF dip at 9–10 kHz, followed by accentuated losses from 13 to 15 kHz and up. For young people 18–24 years old with Grade I–II NIHL, the 3–6 kHz CF dip tends to be accompanied by shifts in the same EHF regions, at 9–10 kHz and above 13–15 kHz. For increasing grades of CF loss, these accentuated EHF shifts characteristically seem to be progressively widened, merging, and being ultimately concealed by more general EHF deterioration, in accordance with reported TTS¹⁰⁵ and PTS¹⁰⁷ data. For older subjects with slight to moderate CF NIHL, and in all age groups for more severe CF loss (Grades III and IV), this EHF deterioration process seems to progress more rapidly. Single events of impulse noise exposure exceeding a critical level may cause extensive CF and EHF hearing loss, also unilaterally, from 2 to 3 kHz and up, as previously reported.^{32,102,109} The accentuated shifts in the EHF range, at 9 kHz and above 13–15 kHz, could hardly be artifacts, as they did not appear in the normative

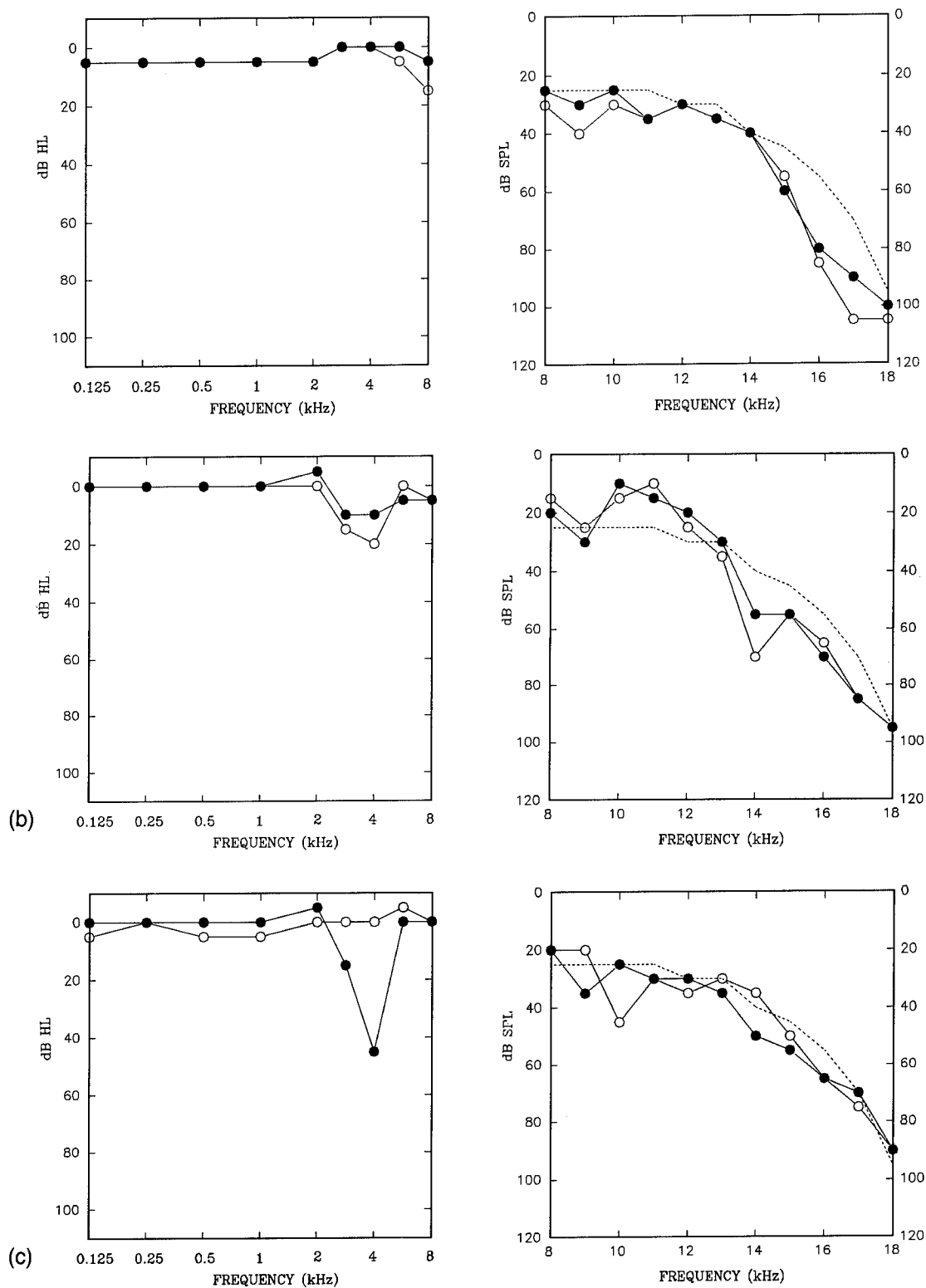


Figure 24-1 Individual air conduction thresholds in the CF and EHF ranges for 10 males, aged: (a–g) 19–22 years; (h) 35 years; (i) 45 years; (j) 54 years. Note that the EHF scale is linear and that the CF ordinate is in dB HL, and the EHF ordinate in dB SPL. (●) Left ear; (○) right ear; (---) normative EHF thresholds from the same laboratory⁴⁰ for the subject's age group.

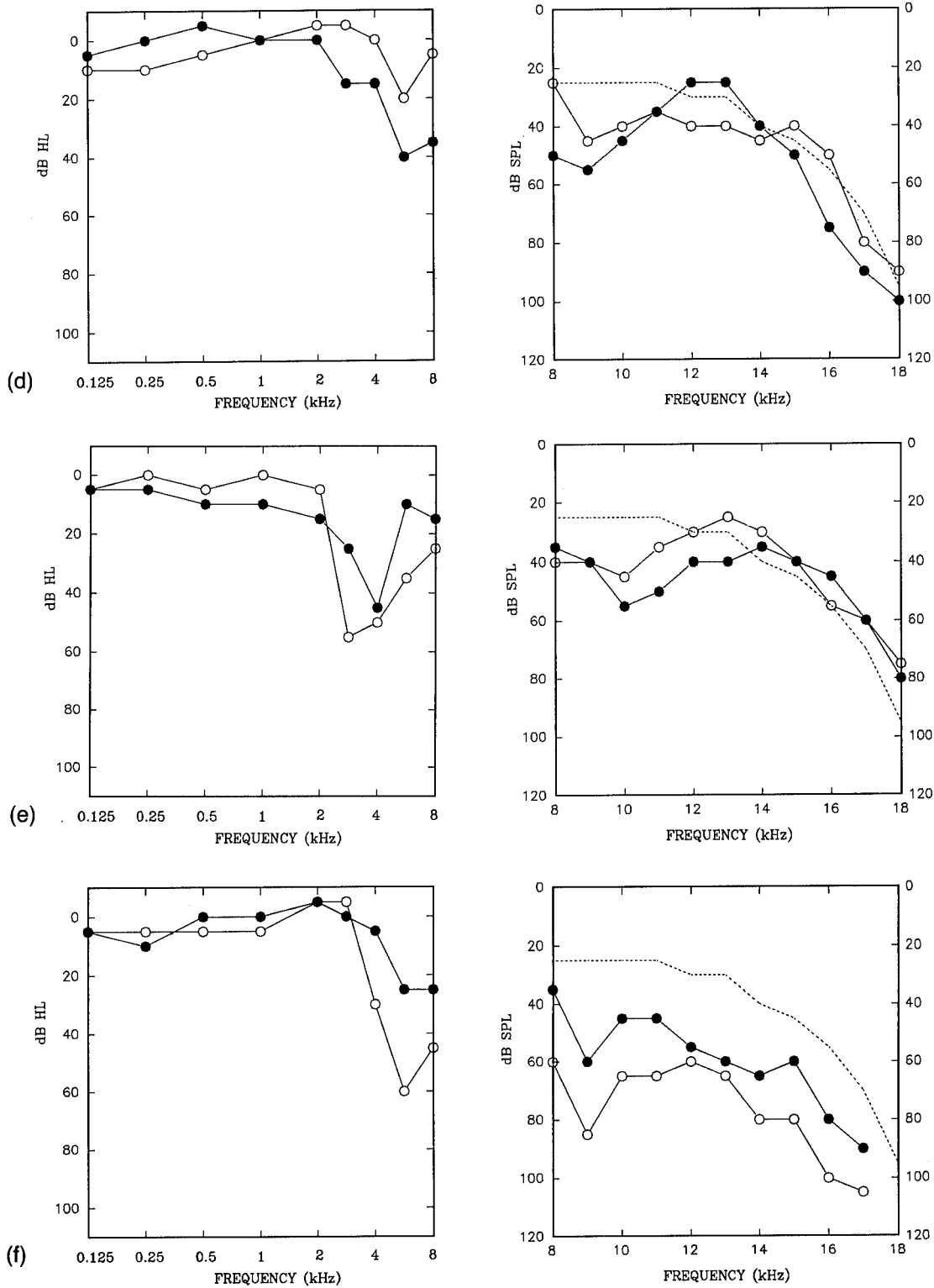


Figure 24-1 Continued

CHAPTER 24 • EXTENDED HIGH-FREQUENCY HEARING LOSS

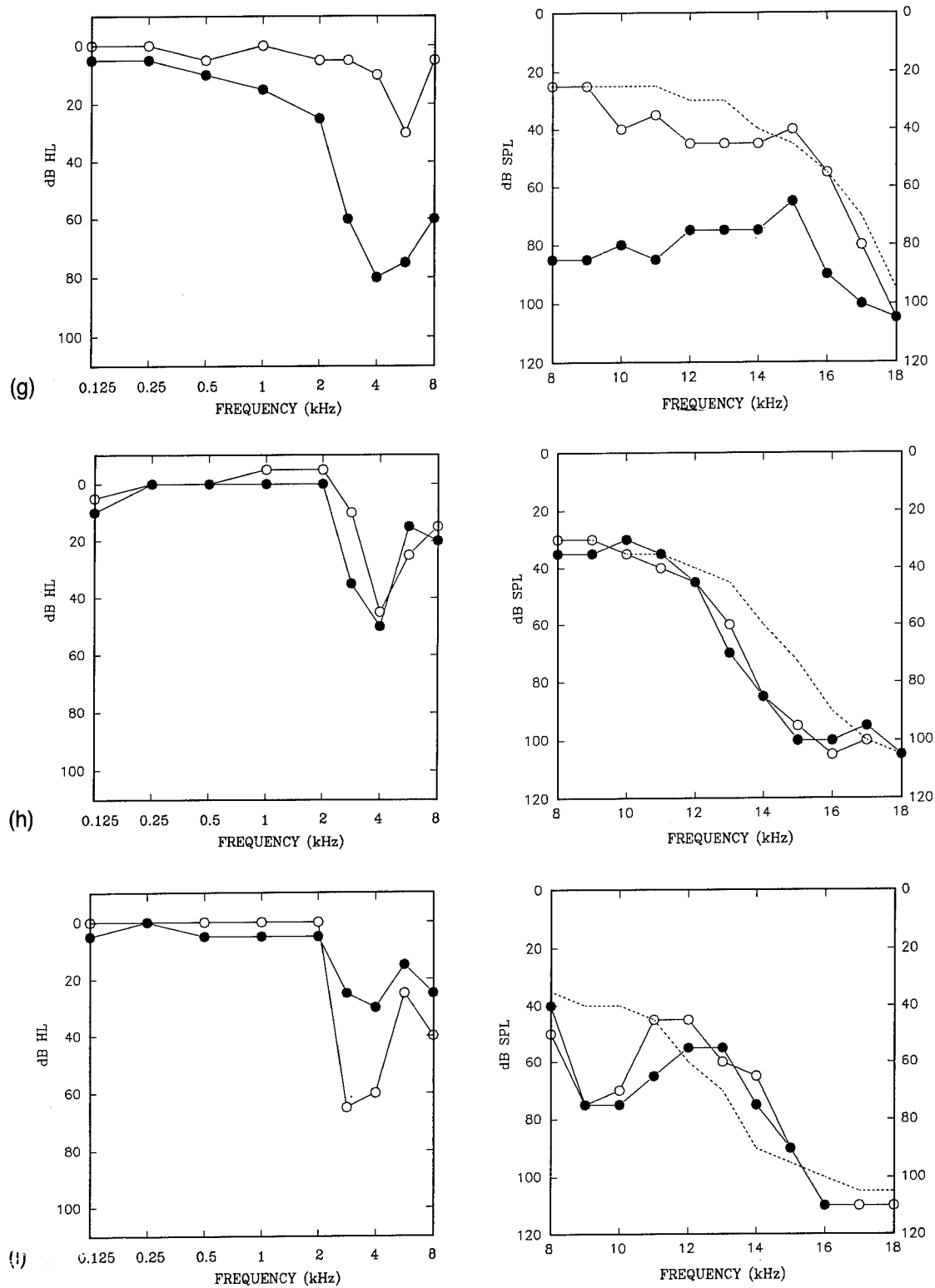


Figure 24-1 Continued

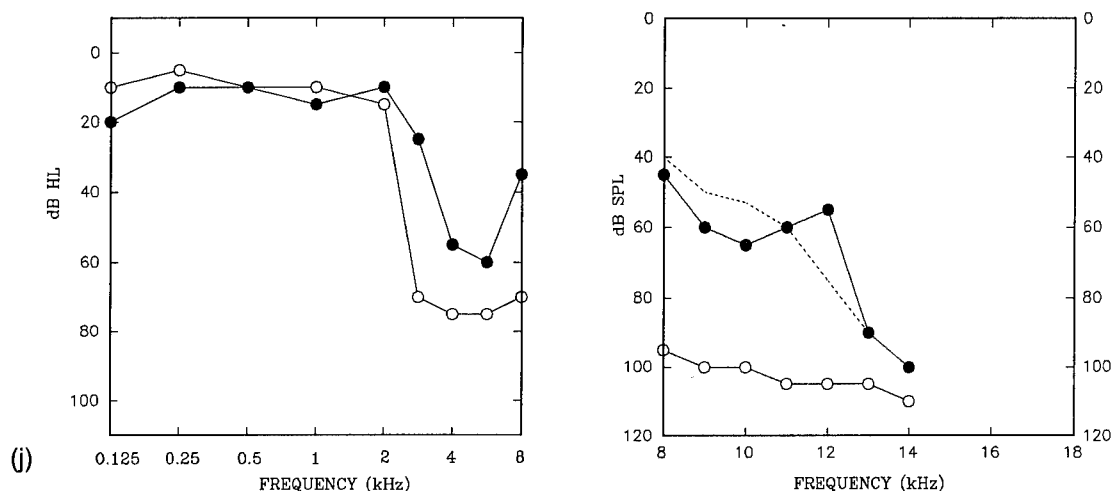


Figure 24-1 Continued

EHF thresholds.⁴⁰ Besides, the EHF shifts increased progressively for the same frequencies with increasing CF loss.

Basilar Membrane Mechanics

In NIHL, morphological damage is found first in the first row of outer hair cells, then in the inner hair cells, then in the second and third row of the outer hair cells.¹¹⁵ A possible pathophysiological mechanism is acoustic trauma-induced swelling of the afferent dendrites below the inner hair cells due to excess glutamate release during mechanical stimulation.¹¹⁶ Protective mechanisms against intense sound seem to be mediated by activation of cochlear dopaminergic lateral efferents against this acoustic trauma-induced swelling, and by medial efferents to reduce the mechanical damage at the outer hair cell level.¹¹⁶ Depending upon the sound level, reversible physiological fatigue (TTS) or irreversible toxic or mechanical damage (PTS) would then be expected to develop at, or related to, movement maxima along the membrane⁸⁷ (see also Chapter 10).

Above a certain critical stimulus level, more extensive, immediate, and permanent mechanical damage occurs, both at the place of the characteristic NIHL "dip," that grows deeper and wider and eventually causes deterioration of the entire high-frequency

region, manifested anatomically as well as audiometrically.^{32,85,102,109,111,113}

The external and middle ear mechanical transfer function accentuates SPL around 2.5–4 kHz.¹¹⁸ Maximum hearing loss is most often detected around 0.5–1 octave above the exposure frequency.^{85–87} Harmonic distortion seems to create octave- and combination-tone-related movement maxima along the basilar membrane, indicated by a number of phenomena involved in pitch perception, consonance, fusion, masking, and two-tone suppression.^{117,118} For moderate SPLs one would then expect TTS to be located at harmonically or octave-related multiples of the stimulus frequency, and/or of the accentuated "transfer function + 0.5–1 octave" frequency, along the basilar membrane as found for TTS in humans by Fritze and Köhler.¹⁰⁵ For levels just sufficient to produce mechanical hair-cell damage, PTS would be expected to develop with maxima at corresponding locations on the membrane. For higher SPLs, increased membrane motion would create mechanical damage also adjacent to the maxima widening the threshold shift dips and damaging the more resistant hair cells at the movement maxima. This in turn would increase the audiometric dips and progressively yield more widespread total damage and an ultimate ceiling effect, in agreement with Ase et al.¹⁰⁷ and the present results.

Age seems to increase the susceptibility and may accelerate the pathophysiological process, or reduce the efficiency of the protective processes.

General Conclusion

Noise exposure leads to elevated EHF thresholds compared with normative thresholds, across age groups and grades of NIHL. For ototoxic and infectious trauma, the EHF shifts seem to be manifested earlier and tend to be more substantial and symmetric than the CF shift, indicating that EHF are more susceptible than CF. Some studies indicate that this is also true for NIHL.

The noise-induced EHF hearing loss tends to be accentuated at around 9 kHz, possibly harmonically or in an octave relation to the characteristic 3–6 kHz CF dip, and simultaneously involves regions above 13–15 kHz. For increasing degrees of CF NIHL, the EHF hearing loss progresses toward more widespread dips which ultimately merge and lead to extensive EHF deterioration until a ceiling effect is reached. Beyond a critical level, single exposures may produce extensive hearing loss from 2–3 kHz and up. Moderate level exposures lead to asymptotic EHF TS. Variability is considerable across individuals, and susceptibility seems to increase with age.

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Chapter 25

Temporary Threshold Shifts Produced by High-Intensity Free-Field Impulse Noise in Humans Wearing Hearing Protection

James H. Patterson, Jr. and Daniel L. Johnson

Over the past several years, the US Army Medical Research and Development Command has sponsored a series of studies to determine the human tolerance limits of exposure to high-intensity free-field impulse noise. These studies have been conducted at the Blast Overpressure Test Site in Albuquerque, NM, by EG&G Management Systems, Inc. The goal of these studies was to provide information relevant to the maximum safe exposure limits for various heavy weapons: towed artillery, mortars, and shoulder fired antiarmor weapons. Pfander¹ reported the results of temporary threshold shift (TTS) studies in which soldiers were exposed to the noise of various weapons. More recently, Patterson et al.^{2,3} and Dancer et al.⁴ reported studies designed to determine TTS in volunteers exposed to artillery and antiarmor weapons. These studies all demonstrated that specific weapons could be fired safely with hearing protection. However, they did not establish new limits for impulse noise exposure because essentially no effects on hearing were found.

In addition to effects on hearing, high intensity blasts can injure other organ systems. The air containing organs seem to be the next most susceptible organs after the inner ears. Dodd et al.⁵ proposed limits for exposure to blasts with minimal risk of upper airway, lung, and gastrointestinal injury. These limits are well above the blast limits in current weapons design standards in the United States.⁶ The

studies reported here were designed to use exposures to impulse levels beyond any that had been used previously in auditory experiments on humans in order to determine the exposures that would produce an effect on hearing. The exposures were limited only by the limits for nonauditory injury.

Methods

The basic approach of the studies was to expose human volunteers to a progression of increasingly more energetic impulse noise stimuli. Hearing protection was worn during all exposures. Temporary changes in hearing threshold (TTS) were used as the basic indicator of adverse effects on hearing. All exposure stimuli were produced by the detonation of high explosives. Three different exposure configurations were used to vary the duration of the impulse by changing the distance between the explosive source and volunteers. The first configuration placed the volunteers 5 m from the detonation. This produced a pressure-time signature (Figure 25-1a) characteristic of towed artillery. The A-duration was approximately 2.9 milliseconds. The second configuration placed the volunteers 3 m from the explosive source. This produced a pressure signature (Figure 25-1b) with a 1.5 millisecond A-duration. The third configuration placed the volunteers within 1 m of the

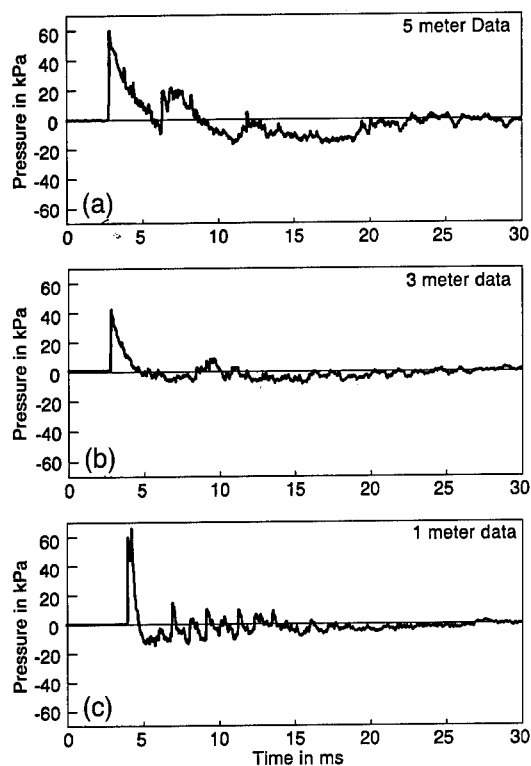


Figure 25-1 Pressure-time signatures at (a) 5 m, (b) 3 m, and (c) 1 m distance conditions.

source. This produced an impulse with a 0.8 millisecond A-duration (Figure 25-1c). Because the A-duration of a free-field impulse strongly influences the distribution of energy across frequency, the three configurations produced exposure stimuli with different energy density spectra. Figure 25-2 shows the spectra of the three impulses. The pressure-time signatures with the longer A-duration have more low frequency energy in the spectrum.

At least 60 volunteers were exposed to impulses at each distance configuration. The primary measure of effect on hearing was TTS immediately after the exposure (2–6 minutes). A criterion of 25 dB TTS was adopted to define unacceptable effects on hearing. The intensity and number of impulses were varied to find the maximum exposure that would produce an unacceptable TTS in 5% of the exposed volunteers. The maximum intensities were set by the nonauditory injury limits derived by

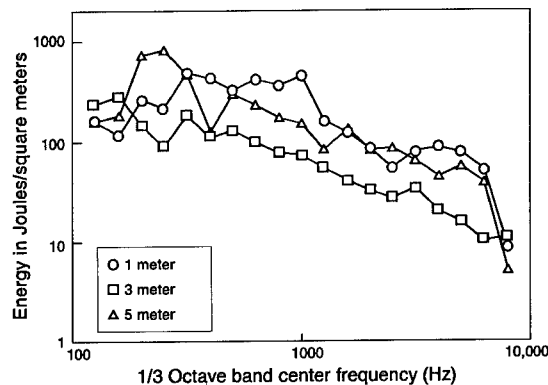


Figure 25-2 One-third octave band spectrum of the three impulses.

Dodd et al.⁵ The number of impulses per exposure was varied from 6 to 100.

Hearing protectors with two different attenuation characteristics were used in these studies. The first protector was an ear muff compatible with the US Army infantry helmet. The attenuation of this hearing protector is shown in Figure 25-3 as the standard muff. It is comparable to other protectors commonly used in the military. The second protector was a modified version of the standard muff. The attenuation was reduced to simulate a poor fit. This was accomplished by inserting plastic tubes through the ear seals to introduce a controlled leak. The attenuation is shown in Figure 25-4 where it is compared to the standard muff.

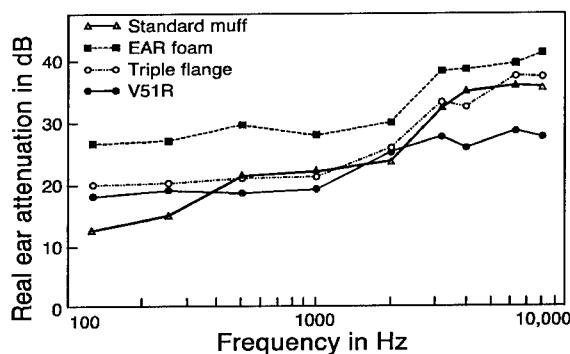


Figure 25-3 Attenuation of the standard earmuff compared to other hearing protectors used by the US Army.

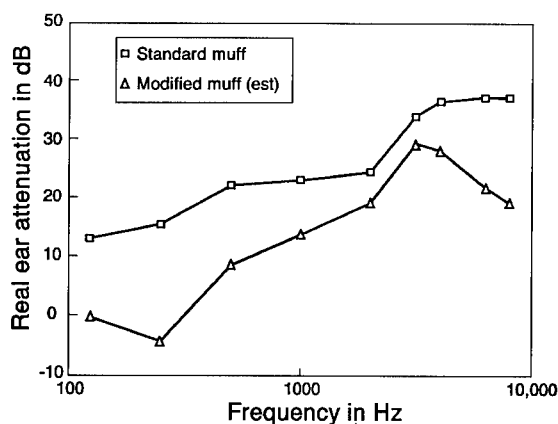


Figure 25-4 Attenuation of the standard earmuff and the modified earmuff.

Results and Discussion

The results of these studies can be summarized as the percentage of volunteers showing unacceptable TTSs (i.e., TTS > 25 dB at any frequency) for each combination of intensity and number of impulses. In addition, we may calculate, using order statistics,⁷ the confidence that no more than 5% of the population would exhibit a TTS exceeding 25 dB. The minimum sample size of 59 volunteers was calculated so that the largest TTS would provide a 95% confidence upper bound on the TTS at the 95th percentile of the population. Thus if the largest TTS did not exceed 25 dB, we can be 95% confident that 95% of the population would not show a TTS larger than 25 dB. The second largest TTS then forms a lower confidence upper bound on the 95th percentile TTS. This sequence may be extended through all the subjects. As a matter of practicality, the confidence drops to approximately 5% at the sixth largest TTS. When 6 out of 59 volunteers show TTSs exceeding 25 dB, we can be 95% confident that the 95th percentile TTS also exceeds 25 dB.

5 m Distance

Two groups of subjects were exposed at the 5 m distance. The exposure levels ranged from 174 to 191 dB peak sound pressure level (SPL). The first group wore the standard earmuff. None of the volunteers exposed at the 5 m

distance with the standard muff incurred a TTS in excess of 25 dB. In fact, none of the volunteers incurred even a 15 dB TTS.

Then, the 5 m exposures were repeated on another group of volunteers wearing the modified muff. This time, TTS in excess of 25 dB was observed in a few volunteers at the most energetic conditions. Figure 25-5 shows the percentage of volunteers showing an unacceptable TTS. Note that even though we started with at least 50 volunteers in each group, the number varied across the studies. Also, as volunteers dropped out of a study, the number of volunteers at each exposure condition within the study varied. Figure 25-6 shows the confidence that 95% of the population would show an acceptable TTS. This incorporates the effects of both the number of volunteers and the number of unacceptable TTSs.

3 m Distance

In the next study, another group of volunteers was exposed at the 3 m distance to intensities ranging from 174 to 193 dB peak SPL with an A-duration of 1.5 milliseconds. The number of impulses per exposure again was varied from 6 to 100. The hearing protection was the modified muff. The most energetic conditions again produced unacceptable TTS in some of the volunteers. Figure 25-7 shows the percentage of volunteers with an unacceptable TTS. In this case, the higher level impulses produced more unacceptable TTSs than at the 5 m distance.

Five volunteers in this group were prevented from proceeding to more energetic conditions because of unusual recovery patterns. These included either recovery times longer than 24 hours or a pattern of growth of TTS during the first 24 hours. The data for these volunteers were included for all conditions in which they participated. As a result, the data in the 25, 50, and 100 shot conditions probably shows fewer unacceptable TTSs than would have occurred if these volunteers had been allowed to continue in the study. Although it is difficult to estimate the effect these volunteers may have had on the data, it is unlikely that they would have reduced the

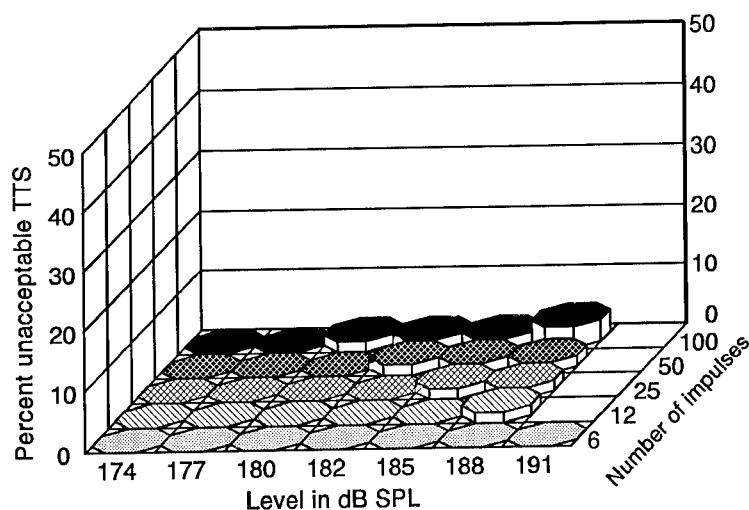


Figure 25-5 Percentage of volunteers showing an unacceptable TTS after exposure at the 5 m distance while wearing the modified earmuff.

maximum safe exposure levels (discussed below) more than 3 dB for 100 shots.

Figure 25-8 shows the confidence that 95% of the population would have a TTS less than 25 dB. These data also are influenced by the discontinued volunteers.

1 m Distance

At the 1 m distance the peak pressures were varied from 178 to 196 dB peak SPL, with

A-durations of 0.8 milliseconds. In this study, the number of impulses per exposure also was varied from 6 to 100 and the volunteers wore the modified muff. Figure 25-9 shows the percentage of volunteers showing a TTS in excess of 25 dB. In this case, five volunteers also were prevented from completing all exposures. Therefore, the comments about potential effects on the data in the 3 m section also apply to the data from this distance. The confidence that 95% of the population exposed to this

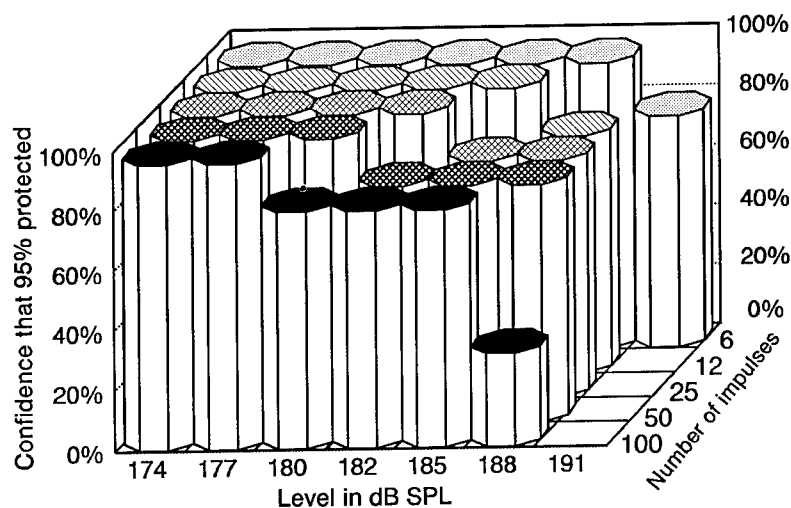


Figure 25-6 Percentage confidence that 95% of the exposed population would show an acceptable TTS after exposure at the 5 m distance while wearing the modified earmuff.

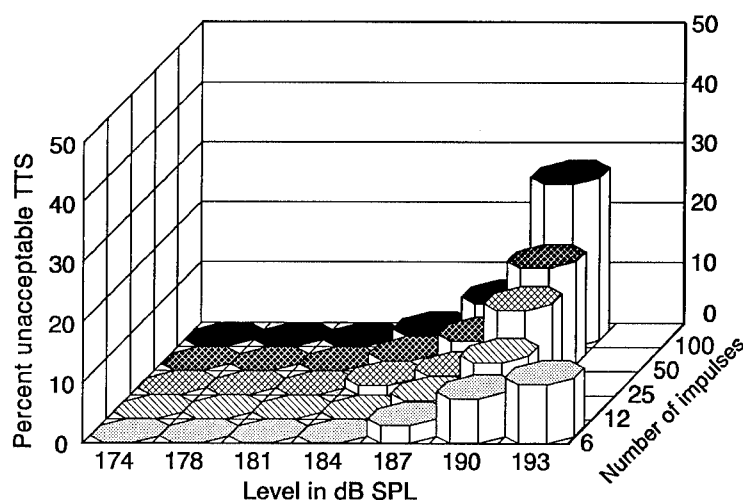


Figure 25-7 Percentage of volunteers showing an unacceptable TTS after exposure at the 3 m distance while wearing the modified earmuff.

impulse would show less than 25 dB TTS is shown in Figure 25-10.

Development of Safe Levels

There are several ways to derive maximum safe exposure levels from the TTS data. Each combination of intensity level and number of impulses defines an exposure condition. One way to estimate the maximum safe exposure levels is to find the set of exposure conditions

for each distance that resulted in unacceptable TTS in <5% of the exposed population (see Figures 25-5, 25-7, 25-9). The maximum safe exposure levels come from the exposure condition with the highest intensity level for each number of impulses for which <5% of the volunteers showed an unacceptable TTS. Table 25-1 contains these levels for all exposure distances.

An alternative way to estimate the maximum safe exposure levels is to use the per-

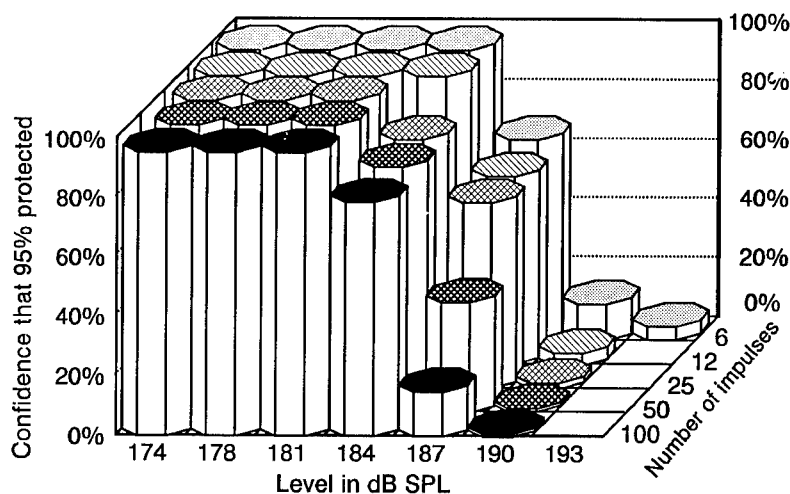


Figure 25-8 Percentage confidence that 95% of the exposed population would show an acceptable TTS after exposure at the 3 m distance while wearing the modified earmuff.

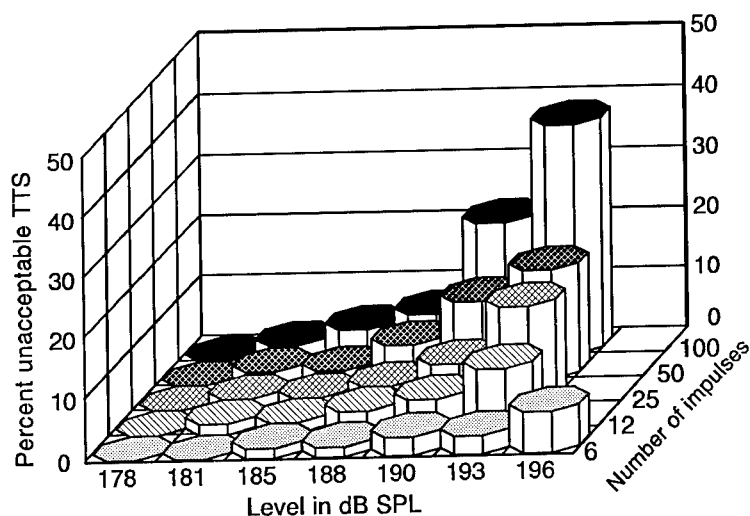


Figure 25-9 Percentage of volunteers showing an unacceptable TTS after exposure at the 1 m distance while wearing the modified earmuff.

centage confidence that 95% of the exposed population would show an acceptable TTS. To do this we must select a percentage confidence to use as defining safe exposure conditions. If we require high confidence (e.g. 95%), the estimated safe levels will be lower. If we choose a low confidence (e.g. 5%), the estimated safe levels will be higher. By choosing a medium value of 50% confidence, we balance these extremes. Then, the maximum safe exposure levels come from the exposure

conditions with the highest intensity level for each number of impulses for which the percent confidence that 95% of the exposed population would show an acceptable TTS is 50% (see Figures 25-6, 25-8, 25-10). These levels are shown in Table 25-2.

As can be seen, there are some differences between these two approaches. However, these differences are no more than one level step in the exposure series used at each distance. Because the number of subjects actually

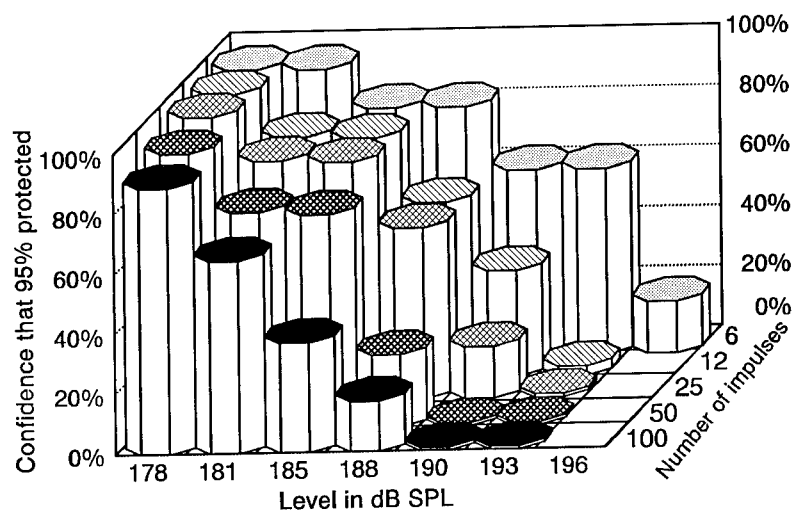


Figure 25-10 Percentage confidence that 95% of the exposed population would show an acceptable TTS after exposure at the 1 m distance while wearing the modified earmuff.

Table 25-1 Maximum Exposure Levels Resulting in at Least 95% Acceptable TTS

No. Impulses	Exposure Condition		
	5 m	3 m	1 m
6	191*	187	193
12	188*	187	190
25	188*	187	188
50	187*	187	185
100	187*	184	185

* Nonauditory limits.

Table 25-2 Maximum Exposure Levels Resulting in 50% Confidence That 95% of Population Show Acceptable TTS

No. Impulses	Exposure Condition		
	5 m	3 m	1 m
6	191*	187	193
12	188*	187	188
25	188*	187	188
50	187*	184	185
100	185	184	181

* Nonauditory limits.

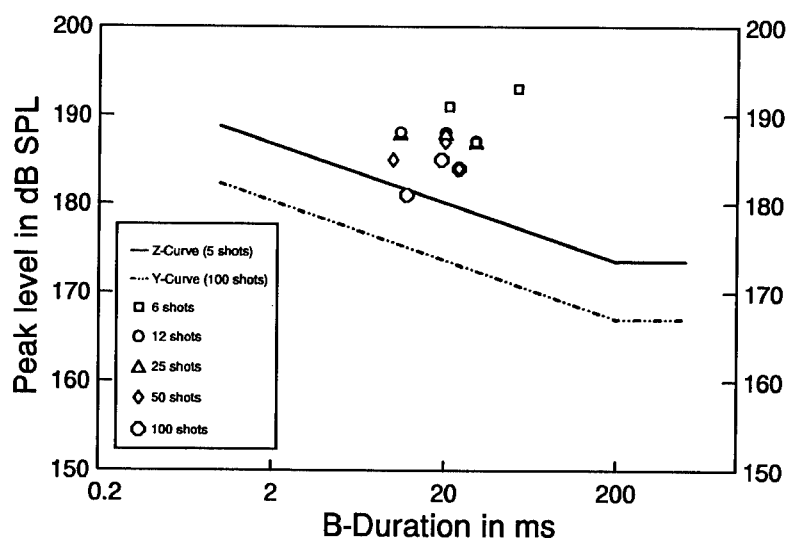
showing an unacceptable TTS was small, these differences are probably statistical fluctuations. The percent confidence incorporates both the number of individuals showing a significant TTS and the statistical effect of the number of volunteers included in each exposure condition. Therefore, it seems reasonable to use the maximum safe exposure levels in Table 25-2.

The values from Table 25-2 are shown in Figure 25-11 with the Z curve (5 shot limit) and the Y curve (100 shot limit) from MIL-STD-1474. Note that the maximum safe exposure levels for various numbers of rounds de-

rived from the studies reported here fall 5–15 dB above the respective limits from the military standard. There also appears to be a trend for the results from this study to slope upward with B-duration while the current Y and Z curve limits from MIL-STD-1474 slope downward with B-duration. This suggests that the peak level and B-duration are not good indicators of auditory hazard.

Conclusions

The results of these studies clearly indicate that an earmuff can provide hearing protec-

**Figure 25-11** Comparison of maximum acceptable exposure levels with US MIL-STD-1474C.

tion for free-field blast levels that greatly exceed our current exposure limits. The use of modified muffs in these studies simulates the commonly occurring situation in which earmuffs do not fit properly, for example, eye glasses temple pieces, long hair, or head gear can compromise the ear seal. Thus, the results should apply to a variety of real world exposure situations. Therefore, we may conclude from these studies that even poorly fitting earmuffs can provide adequate protection against heavy weapons noise in the range of 181–194 dB peak SPL.

While the results of these studies clearly show that the current military exposure limits are too restrictive, the replacement limits are not yet defined. The new limit for free-field impulses will, in all likelihood, depend on the spectrum of the impulse, the attenuation characteristic of the hearing protector, and the number of impulses. How these factors will interact to produce the exact exposure limits is still being explored.

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Chapter 26

Hearing Protector Performance and NIHL in Extreme Environments: Actual Performance of Hearing Protectors in Impulse Noise/Nonlinear Behavior

Armand L. Dancer, Rodolphe Franke,
Georges Parmentier, and Karl Buck

In some circumstances, people can be exposed to very high-level noises: continuous noises in the vicinity of jet engines can easily exceed 130 dBA and impulse noises produced by weapons can reach 190 dB peak at the ear! Although these extreme exposure conditions are relatively infrequent and concern only a few people, they present serious problems because they can produce immediate cochlear lesions and hence, large permanent threshold shifts (PTS).¹ Moreover, even "moderate" intensity weapon noises (150–165 dB peak) such as those produced by rifles either in military training or during recreational activities (target shooting, hunting, etc.)² correspond to a major cause of acoustic trauma among the young male population.^{3,4} Subjects exposed to these noises must be equipped with correctly fitted single (or double) hearing protectors (HP) offering adequate performance. The present study aims to determine the best way to measure the physical attenuation afforded by HP even in extreme acoustical environments.

For the assessment of the attenuation afforded by earplugs and earmuffs at very high-level impulse noises, the classical measurements performed by means of the subjective real-ear attenuation at threshold (REAT) method at low steady-state noise levels (according to ISO 4869-1, for example)⁵ are not

suitable. First of all, this method does not allow an evaluation of the peak pressure of an impulse under an HP. Currently the ISO standard restricts the equal-energy principle to peak levels below 140 dB (ISO 1999)⁶; the American Conference of Governmental Industrial Hygienists (1988)⁷ does not allow the unprotected exposure to impulses above 140 dB peak; and most damage risk criteria (DRC) for weapon noises¹ take into account, besides the duration and the number of the impulses, the peak pressure.^{8,9} Even if serious doubts exist about the pertinency of "peak level" measurements under HP as part of the classical DRC (these DRC actually settle limits for unprotected ears and refer to measurements performed in the free field^{10–12}), it is nevertheless essential to get this information. Moreover, the behavior of the HP undergoing the action of large impulses may exhibit nonlinearities. The appearance of a nonlinearity, its importance, the variation of its characteristics as a function of the pressure–time history of the impulses, as well as its net effect (either "positive" or "negative" as far as the global attenuation is concerned) are generally unpredictable. For this reason the attenuation of each HP should be measured in the actual exposure conditions for which it is intended to be used. The same kind of limitations apply to the "microphone in real ear" (MIRE) measurement

technique.¹³ In practice the peak level and pressure-time history of the impulses cannot be easily and directly measured close to the tympanum by MIRE. Moreover this technique is presently unsuitable for earplug attenuation measurements and is impossible to use as a routine technique with high-intensity impulses because of the security of the subjects.

Therefore, the only possibility to assess the actual behavior of an HP when exposed to impulse noises (up to 190 dB sound pressure level, SPL), to characterize their nonlinearity (if any), and to measure the amplitude spectrum and peak pressure attenuation, is to use an acoustical test fixture (ATF) and preferably an artificial head with an ear simulator.^{14,15}

ATFs

ATFs are currently used to evaluate the physical attenuation afforded by earmuffs in steady-state noise. In these conditions the ATF must comply with ISO and/or ANSI standards^{16,17} (Figure 26-1).

The Brüel & Kjaer ATF (type 4128) is equipped with an ear simulator (type 4157). Because the internal structure of this ATF (i.e.,

mouth simulator) does not allow easy modification of the measuring equipment, it was not used in our study.

The Knowles electronic manikin for acoustic research (KEMAR) ATF¹⁸ is equipped with a Zwislocki ear simulator¹⁹ whose acoustic isolation is poor at low frequencies.²⁰ This ATF has been radically modified and improved by Kunov and Giguère²¹: better isolation (Figure 26-1), flesh simulation in the circumaural region and in the ear canal, compliant neck, etc. Giguère and Kunov²² performed HP attenuation measurements under steady-state and impulse noise conditions (up to 134 dB peak) and stressed the fact that simulation of the auditory canal tissues is essential in the case of measurements with earplugs. However, Kunov and Giguère's ATF is not commercially available.

The HEAD Acoustics GmbH ATF is of recent design. It is equipped with the Brüel & Kjaer ear simulator (type 4157) and incorporates flesh simulation of the circumaural region and of the ear canal.

We have tested the acoustic isolation of the KEMAR ATF and of the HEAD Acoustics GmbH ATF (first version) when exposed to

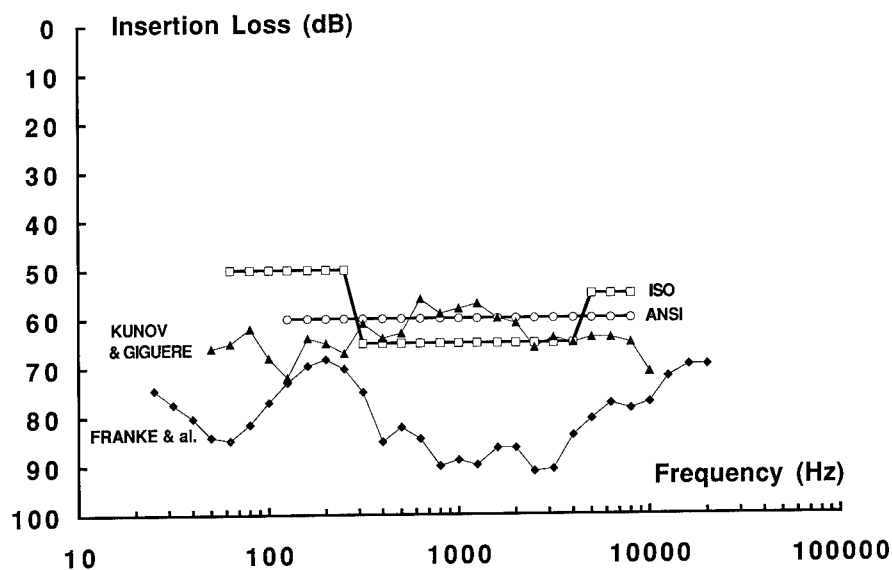


Figure 26-1 Minimal insertion loss of ATF (dB) as defined by the (□) ISO and (○) ANSI standards. (▲) Insertion loss performance of the KEMAR ATF as modified by Kunov and Giguère²⁰; (◆) insertion loss performance of the ISL ATF¹³ ($1/3$ -octave bands).

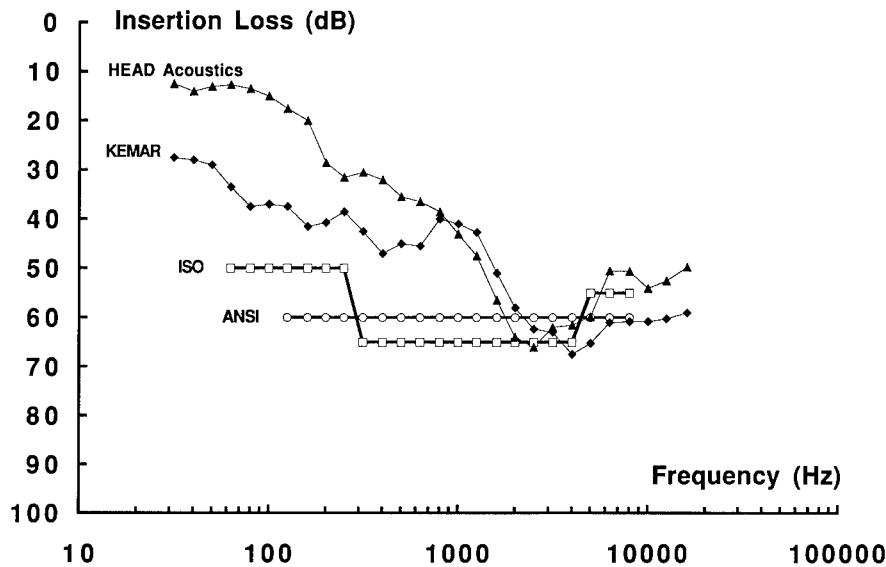


Figure 26-2 Insertion loss performance of the (◆) KEMAR and (▲) HEAD Acoustics (first version) ATFs. Minimal insertion loss of ATF (dB) as defined by the (□) ISO and (○) ANSI standards ($1/3$ -octave bands).

impulses produced by the detonation of bare charges of explosive in the free field (Friedlander waves). The ATFs were equipped with double protection (Willson SB 258F earmuff with a metallic plug tightly closing the entrance to the ear simulator) to determine the maximum insertion loss (IL) that can be measured.²³ Because we wanted to perform IL measurements in excess of 60 or 70 dB over a broad frequency range (the interpretation of such physically measured IL values for the actual protection of hearing will be discussed later), the results (Figure 26-2) were not judged satisfactory (Berger²⁴ came to the same conclusion concerning the KEMAR ATF).

The only solution left was to design a new ATF in order to obtain better performance characteristics (Figure 26-3). The "head" was made of beechwood and was arranged to fit the HEAD Acoustics GmbH device corresponding to the external ear and the circumaural region, and the Brüel & Kjaer ear simulator (type 4157). A brass shell surrounds the ear simulator and the preamplifier (type 2633). This shell is suspended inside the head with the help of a spring and special damping foam. To allow the measurement of peak

levels up to 190 dB, the 0.5 in. Brüel & Kjaer microphone (type 4134) of the original ear simulator is replaced by an underpolarized (28 V instead of 200 V) 0.25 in. microphone (type 4136). This modification alters neither the volume of the artificial "tympanic cavity" nor the acoustic impedance of the ear simulator. The transfer function of the open ear (TFOE) of this ATF was measured under grazing and normal incidence with Friedlander waves (peak pressures, 149–179 dB; A-duration, 0.2–2.2 milliseconds) (for a Friedlander wave, the A-duration corresponds to the duration of the first positive phase of the impulse²⁵). The measured TFOE are in close agreement with the experimental data published by Shaw²⁶ and do not depend on the peak pressure or on the duration of the impulses. The TFOE of the ATF can be considered linear up to a peak pressure of about 190 dB at the ear simulator microphone.

When the ATF "ear canal" is closed by two plugs inserted one after the other (first a metallic plug is inserted into the metallic artificial ear canal of the Brüel & Kjaer ear simulator and, then an E·A·R foam earplug (without any coating) is put into the rubberlike artificial ear

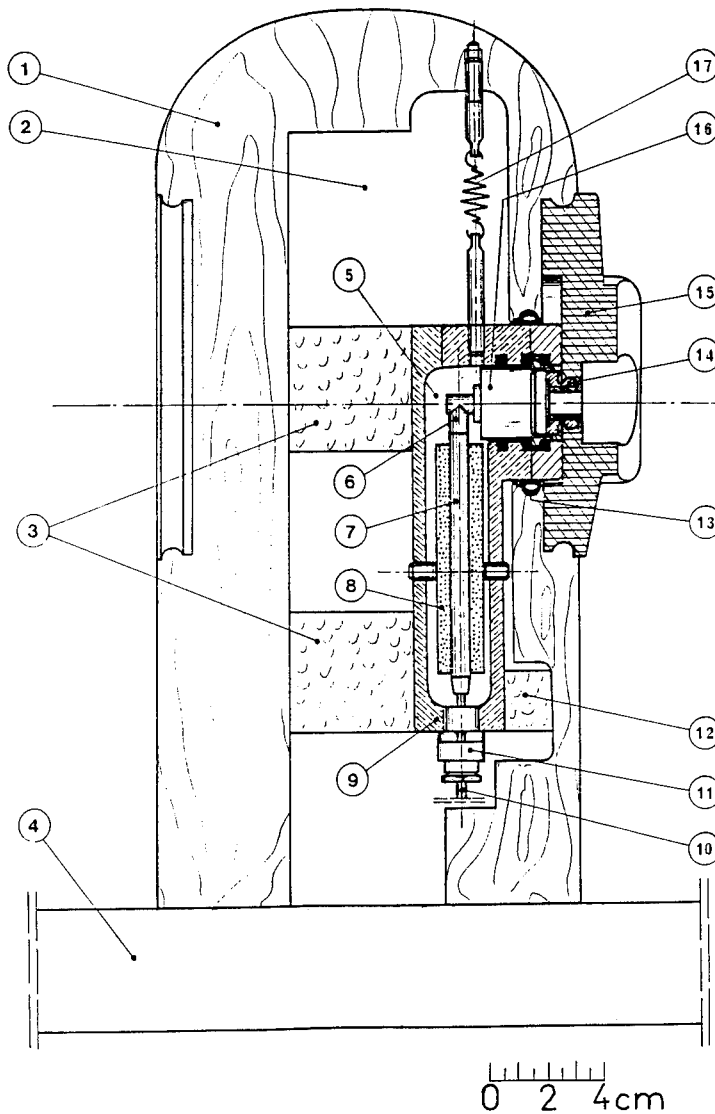


Figure 26-3 Cross section of the ISL ATF: 1. beechwood shell; 2. head cavity; 3. damping foam blocks; 4. wood base; 5. cavity of the brass shell surrounding the measuring equipment; 6. B&K bend (type WU 0278); 7. microphon preamplifier (B&K 2633); 8. foam sleeve; 9. brass shell; 10. cable; 11. stuffing box; 12. damping foam block; 13. circular coupling; 14. auditory canal extension (HEAD Acoustics); 15. outer ear cheek (HEAD Acoustics); 16. ear simulator (B&K 4157) equipped with a B&K microphone (type 4136) and a B&K adaptor (0.5–0.25 in.); 17. suspending spring.

canal of the HEAD Acoustics GmbH device)* the maximum IL afforded by the ATF when

*When the E·A·R foam earplug is put into the rubberlike artificial ear canal alone, that is, without the metallic plug, the attenuation obtained during exposure of the ATF to a continuous noise (70 dB) is: 0.125 kHz, 26 dB; 0.25 kHz, 33.5 dB; 0.5 kHz, 47 dB; 1 kHz, 44 dB; 2 kHz, 43 dB; 4 kHz, 66 dB; 8 kHz, 54 dB.

exposed to Friedlander waves under normal incidence (peak pressure, 162 dB; A-duration, 2.7 milliseconds) is better than 80 dB from 0.4 to 5 kHz (Figure 26-1) and well over the ANSI and ISO criteria.

It is now possible to study the IL of HP without any limitation on the dynamics of the measurements. This ATF is perfectly suitable

for measurements with earmuffs. Concerning the earplugs, although the HEAD Acoustics device provides some simulation of the "ear canal" tissues this feature still needs to be improved: that is, thickness, compliance, geometry, etc. and would require international standardization. Presently our IL measurements for earplugs cannot be regarded as final. Moreover, as the mechanical behavior of some earplugs (i.e., foam plugs) depends largely on their temperature, it is necessary to control the temperature of the "ear canal" walls and to stabilize it around its physiological value.

Insertion Loss Characteristics of Hearing Protectors in High-Level Impulse Noise

The attenuation afforded by seven earplugs (E·A·R foam®, E·A·R Ultrafit®, E·A·R Ultra-tech®, E·A·R Link®, RACAL Airsoft®, RACAL Gunfender®, and a perforated earplug prototype) and 2 earmuffs (WILLSON SB 258® and E·A·R Ultra 9000®) was measured during exposure to Friedlander waves of ≈ 150 , 170, and 190 dB peak pressures (A-durations, ≈ 0.2 and ≈ 2 milliseconds) for normal and grazing incidence.¹⁵ The typical pressure-time history as well as amplitude spectra of these impulses are presented in Figure 26-4. Only the results corresponding to the representative behavior of the HP as observed during this study, will be presented in the following.

Linear Earplug

Figure 26-5 represents the pressure-time histories of the impulses recorded in free field (≈ 191 dB peak; A-duration, ≈ 2 milliseconds, grazing incidence) (Figure 26-5a), and under the RACAL Airsoft earplug (Figure 26-5c). The shape of the pressure recorded under this earplug is quite simple and looks like the pressure time-history in the free field (with the exception of the longer rise-time and the global attenuation). However, this is not always the case. Figure 26-5b represents the same type of recordings for another (quasi-linear) earplug (E·A·R foam). One can notice a

very large negative phase. These observations cast serious doubt on the validity and the actual value of the peak pressure measurement under the HP in the evaluation of the hearing hazard from impulses (as proposed by Pekkarinen et al.⁹).

Figure 26-6 presents the RACAL Airsoft IL as a function of frequency ($1/3$ -octave bands) for impulses of ≈ 150 , 170, and 190 dB peak (≈ 2 milliseconds A-duration, normal incidence). No significant modification of the IL is observed when the peak pressure of the impulse changes. Hence, this earplug is quasi-linear in our experimental conditions. It is interesting to notice (Figure 26-6) that the attenuation of the RACAL Airsoft as measured by REAT methods (ISO 4869 and ANSI S 3.19-1974) at low steady-state noise levels by the Berufsgenossenschaftliches Institut für Arbeitssicherheit²⁷ is comparable to our results obtained with high-level impulses. This indicates clearly that this earplug behaves linearly and that our ATF reproduces the average behavior of the human ear reasonably well.

For given exposure conditions, the reproducibility of our results is almost perfect for a given positioning of the earplug (standard deviation, SD, ≤ 1 dB from 30 to 16 kHz). A larger variability is observed when the earplug is removed and put into place again by the same operator after each exposure (SD reaches 5 dB, at most, between 4 and 10 kHz).¹⁵

Nonlinear Earplug

The pressure-time signatures of impulses recorded in the free field (≈ 191 dB peak; A-duration, ≈ 2 milliseconds, grazing incidence), and under the RACAL Gunfender earplug are presented in Figure 26-5d (note the large negative pressure phase). The RACAL Gunfender earplug is made intentionally nonlinear by means of a metallic plate (0.15 mm thickness) inserted inside the earplug perpendicular to the plug axis and perforated in its center by a hole (0.5 mm diameter). This earplug has been proven to act as a nonlinear mechanism allowing the attenuation values to increase with the stimulation level beyond 120–140 dB.^{28,29}

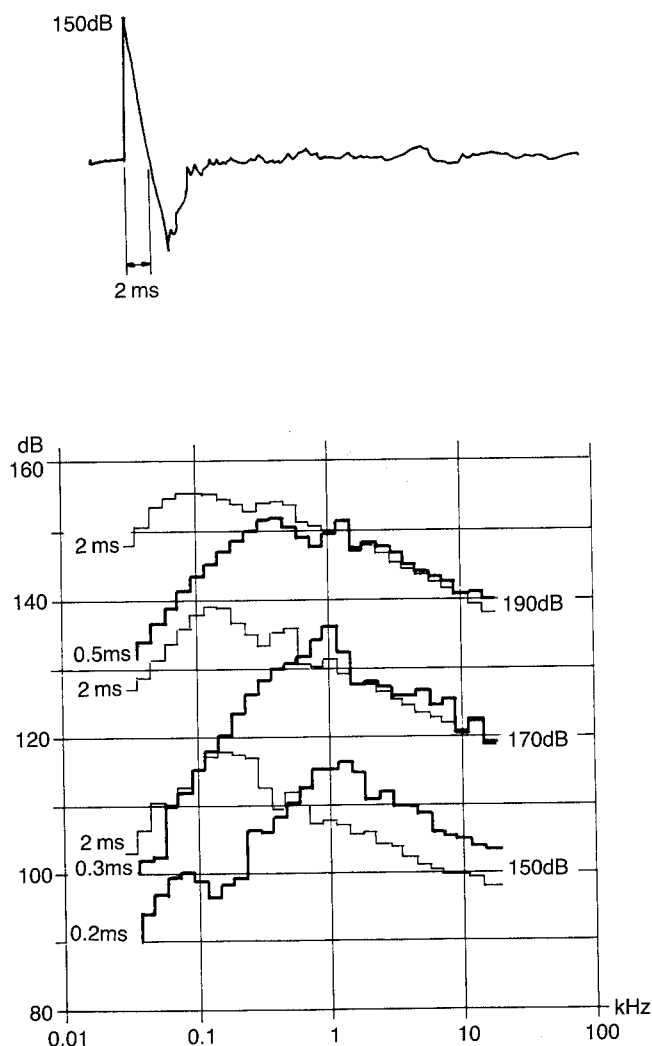


Figure 26-4 Pressure time signature of a Friedlander wave (peak pressure, 150 dB; A-duration, 2 milliseconds) and $1/3$ -octave band amplitude spectra of typical impulses (peak pressure, 150, 170, and 190 dB peak pressure; A-durations, 0.2–2 milliseconds).

Figure 26-7 presents the IL of the RACAL Gunfender earplug as a function of frequency ($1/3$ -octave bands) for impulses of ≈ 150 , 170, and 190 dB peak (2 milliseconds A-duration, grazing incidence). The IL increases by about 10 dB for each 20 dB increase of the peak pressure of the impulse. As a comparison, the IL measured on our ATF at low level (70 dB) steady-state noise, and by the REAT method (ANSI S 3.19-1974) (E.H. Berger, unpublished data, 1993) are presented in the same figure. In this case it is particularly obvious that the

actual attenuation provided at high levels cannot be inferred from REAT attenuation values. The nonlinearity of this earplug is highly favorable, that is, the higher the peak pressure of the impulses, the higher the IL.

Earmuffs

Figure 26-8 presents the pressure–time signatures of the impulses recorded in the free field (Figure 26-8a) (≈ 188 dB peak; A-duration, ≈ 2 milliseconds, grazing incidence) and under

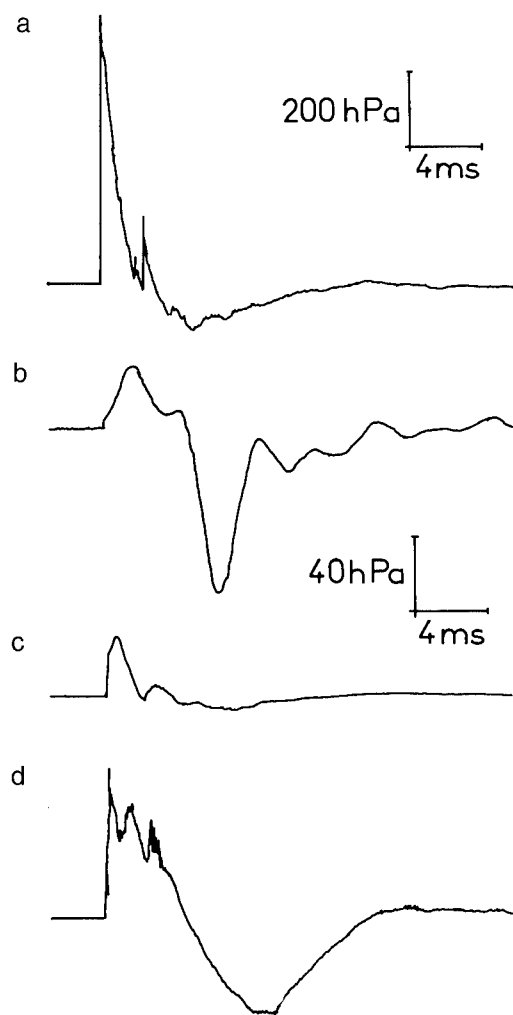


Figure 26-5 Pressure time signatures of a Friedlander wave (grazing incidence) of 191 dB peak and 2 ms A-duration recorded in (a) the free field and under three earplugs: (b) E·A·R foam, (c) RACAL Airsoft, and (d) RACAL Gunfender.

the WILLSON SB-258 earmuff (Figure 26-8b). Here too one can notice a very large negative phase under the HP. Figure 26-9 presents the IL of this muff as a function of frequency ($1/3$ -octave bands) for impulses of ≈ 150 , 170, and 190 dB peak (≈ 2 milliseconds A-duration, normal incidence). Beyond 170 dB, the HP becomes nonlinear: at 190 dB the IL decreases by about 10 dB between 0.3 and 3 kHz. The characteristics of the nonlinearity change in a complicated way as a function of the duration of the impulses and of the angle of incidence.

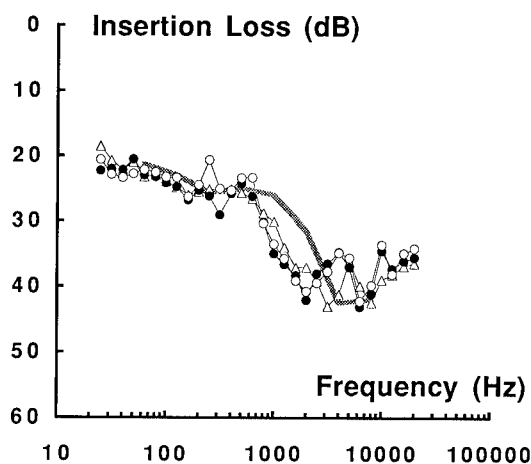


Figure 26-6 Insertion loss afforded by the RACAL Airsoft earplug for different peak pressure levels of the impulses: (○) 150 dB, (●) 170 dB, and (△) 190 dB (A-duration, 2 milliseconds, normal incidence). (hatched line) REAT insertion loss measured by Berufsgenossenschaftliches Institut für Arbeitssicherheit.²⁶ ($1/3$ -octave bands).

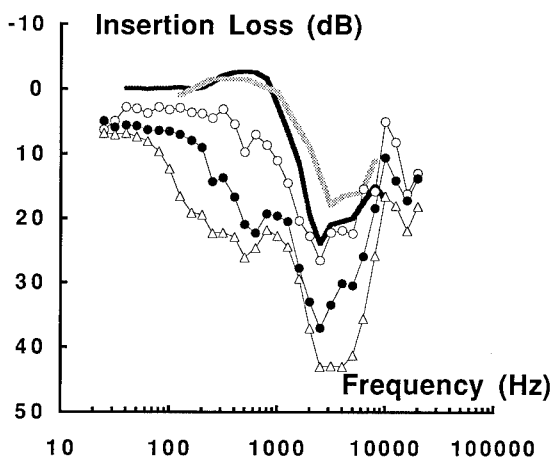


Figure 26-7 Insertion loss afforded by the RACAL Gunfender earplug for different peak pressure levels of the impulses: (○) 150 dB, (●) 170 dB, and (△) 190 dB (A-duration, 2 milliseconds, grazing incidence). (—) Insertion loss measured with the help of the ISL ATF at low-level steady-state noise (70 dB). (hatched line) Insertion loss measured with the help of REAT method by E.H. Berger (unpublished data, 1993). ($1/3$ -octave bands).

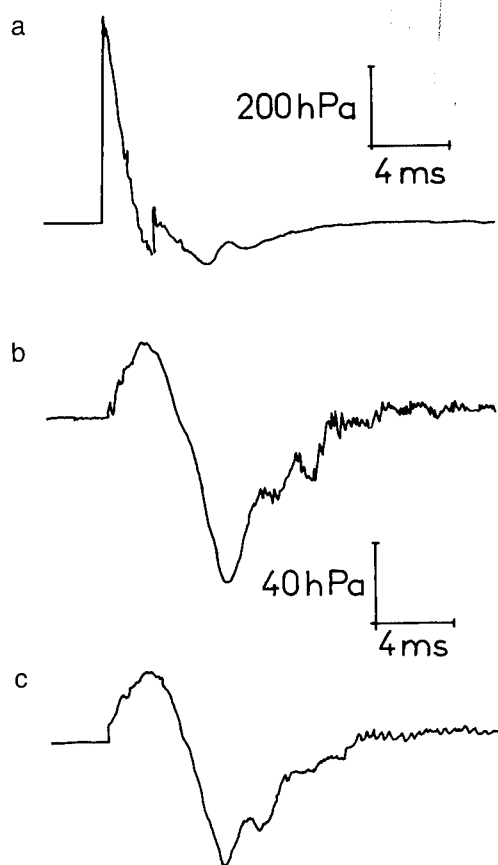


Figure 26-8 Pressure-time signatures of a Friedlander wave (grazing incidence) of 188 dB peak and 2 milliseconds A-duration recorded (a) in the free field, (b) under the WILLSON SB 258, and (c) under the E·A·R Ultra 9000.

However, these changes correspond to an unfavorable nonlinearity: the higher the peak pressure, the lower the IL.

The free-field impulses shown in Figure 26-8a yield the pressure-time history under the E·A·R ULTRA 9000 earmuff shown in Figure 26-8c (note the large negative phase). This entirely passive muff is designed to be nonlinear and to present larger attenuation values when the noise level increases beyond 110 dB. Figure 26-10 presents the IL of this muff as a function of frequency ($1/3$ -octave bands) for impulses of ≈ 150 , 170, and 190 dB peak (≈ 2 milliseconds A-duration, normal incidence). The behavior of this muff is rather complex. It appears that there is a favorable nonlinearity

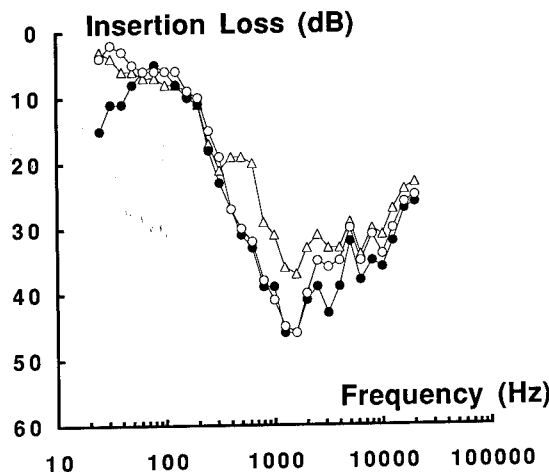


Figure 26-9 Insertion loss afforded by the WILLSON SB 258 earmuff for different peak pressure levels of the impulses: (○) 150 dB, (●) 170 dB, and (△) 190 dB (A-duration, 2 milliseconds, normal incidence) ($1/3$ -octave bands).

from 2 to about 10 kHz, IL increases when the level of the impulse goes up from 150 to 190 dB; and an unfavorable nonlinearity below 2 kHz, IL decreases beyond 170 dB. The comparison with the IL measured either with our ATF or with the REAT method (see Figure

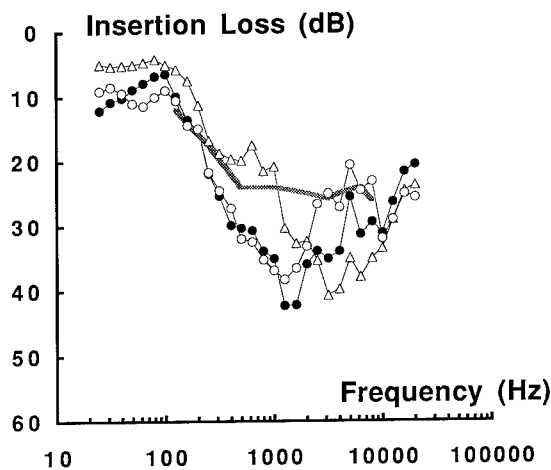


Figure 26-10 Insertion loss afforded by the E·A·R Ultra 9000 earmuff for different peak pressure levels of the impulses: (○) 150 dB, (●) 170 dB, and (△) 190 dB (A-duration, 2 milliseconds, normal incidence). (hatched line) Insertion loss measured with the help of REAT method by E.H. Berger (unpublished data, 1993) ($1/3$ -octave bands).

26-10, according to E.H. Berger, unpublished data, 1993) at low steady-state noise levels indicates, however, a global trend of favorable nonlinearity.

Double Hearing Protection

Two HP (E·A·R foam earplug and WILLSON SB 258 earmuff) were used in combination on our ATF. Figure 26-11a represents the pressure-time signature of the impulse recorded in the free field (≈ 190 dB peak; A-duration, ≈ 2 milliseconds, grazing incidence), and Figure 26-11b the pressure time history under the double hearing protection. Figure 26-12 presents the IL corresponding to this double protection as a function of frequency ($1/3$ -octave bands) for impulses of ≈ 150 , 170, and 190 dB peak (≈ 2 milliseconds A-duration, grazing incidence). It is interesting to notice that in this case and in spite of the presence of the WILLSON SB 258 earmuff that has been shown to behave in a nonlinear way, the double hearing protection behaves linearly (this is

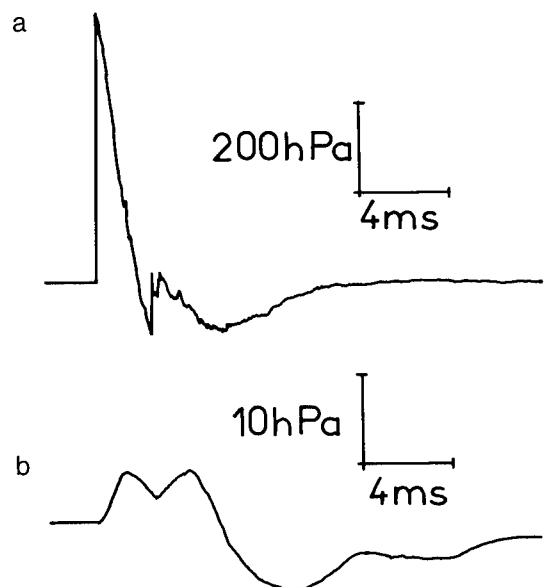


Figure 26-11 Pressure-time signatures of a Friedlander wave (grazing incidence) of 190 dB peak and 2 ms A-duration recorded (a) in the free field and (b) under a double HP: WILLSON SB 258 earmuff plus E·A·R foam earplug.

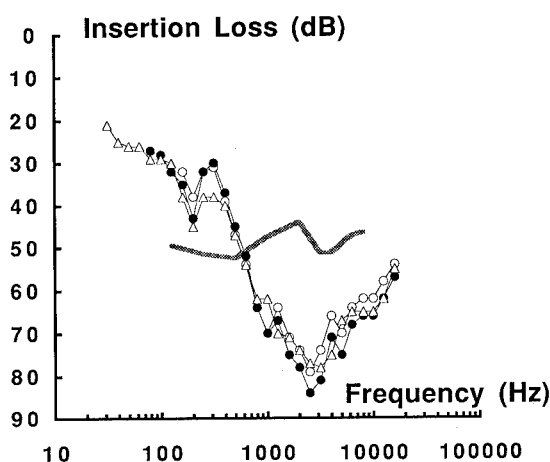


Figure 26-12 Insertion loss afforded by a double HP: WILLSON SB 258 earmuff and E·A·R foam earplug for different peak pressure levels of the impulses: (○) 150 dB, (●) 170 dB, and (△) 190 dB (A-duration, 2 milliseconds, grazing incidence). (hatched line) Average bone conduction limits (according to Schroeter and Els²⁹ and Berger³⁰).

confirmed by comparing these results to the IL measured on the ATF for the low-level steady-state noise of 70 dB). The IL is larger than 70 dB from 1.5 to 4 kHz.

Generally speaking such large IL values are not taken into account for hearing protection evaluation because they exceed the bone conduction (BC) thresholds (see Figure 26-12) (mean values of the BC limits according to Schroeter and Els³⁰ and Berger³¹). However, the measurements that are feasible because of the large dynamic range of our ATF allow: a determination of the physical performance of (almost) any HP; a knowledge of the actual pressure-time history existing below an HP under (almost) any exposure conditions; and an application of any correction curve corresponding to either the BC limits, or to the physiological masking noise (PM) and the occlusion effect (OE),³² and thus allow a very general approach to the measurement of the HP efficiency. Moreover, it must be noted that all investigations of the BC limits rely on threshold detection methods and that no proof exists to indicate that BC noise-induced hearing loss (NIHL) is comparable to that produced by aerial conduction.

Peak Pressure Attenuation and L_{Aeq8} Attenuation

When our ATF is equipped with an HP, the signal measured by the microphone of the ear simulator can be subtracted from the signal measured in the free field (see for example Figures 26-5, 26-8, and 26-11) to calculate: the noise reduction (NR) spectrum; the attenuation of the peak pressure; or the attenuation of the L_{Aeq8} (generally speaking the NR can also be obtained by subtracting the transfer function of the open ear from the insertion loss measurements: $NR \approx IL - TFOE$).²³

The peak pressure attenuation and the L_{Aeq8} attenuation afforded by the earplugs, the earmuffs, and the double protection when exposed under normal incidence to impulses of ≈ 150 , 170, and 190 dB (A-duration, ≈ 2 milliseconds) are reported in Table 26-1 (NR

Table 26-1 Peak Pressure Attenuation and L_{Aeq8} Attenuation (dB)

Δp (ff)	Att. Peak	L_{Aeq8} (ff)	Att. L_{Aeq8}
<i>RACAL Airsoft earplug</i>			
150	22	67	22
170	23	88	22
190	22	108	22
<i>RACAL Gunfender earplug</i>			
150	3	67	3
170	10	88	12
190	15	108	16
<i>WILLSON SB 258 earmuff</i>			
150	14	67	15
170	14	88	17
190	15	108	16
<i>E·A·R ULTRA 9000 earmuff</i>			
150	18	67	19
170	17	88	20
190	12	108	15
<i>Double Protection: E·A·R foam earplug + WILLSON SB 258 earmuff</i>			
150	37	67	34
170	35	88	34
190	36	108	38

Attenuations afforded by earplugs, earmuffs, and double protection when exposed to high-level impulses (peak pressure ≈ 150 , 170, and 190 dB; A-duration ≈ 2 milliseconds, normal incidence) in free field (ff).

values). As can be seen, the peak pressure and L_{Aeq8} attenuation do not depend significantly on the level of the impulses for the RACAL Airsoft earplug and the WILLSON SB258 earmuff. On the other hand, for the RACAL Gunfender earplug the attenuation values increase (by about 12 dB) when the level of the impulses increase from 150 to 190 dB, whereas they decrease by 5 dB for the E·A·R ULTRA 9000 earmuff when the level of the impulses increase from 170 to 190 dB. (Note: this 5 dB decrease is obtained from measurements performed on a single example of this earmuff and cannot be considered as representative of the average behavior of the model.) We can observe in Table 26-1 that the attenuation figures are nearly the same for the L_{Aeq8} as for the peak pressure. This similarity is peculiar to this type of impulse (Friedlander waves of about 2 milliseconds A-duration) and cannot be generalized. The attenuation values reported for each HP, either for the L_{Aeq8} or for the peak pressure, depend on the type of impulses, that is, when the A-duration is shorter (≈ 0.3 milliseconds), the attenuation values of the L_{Aeq8} are generally smaller than those of the peak pressure. Consequently, the attenuation values reported in Table 26-1 must not be regarded as typical of the HPs that have been studied. For an impulse of 150 dB peak, when the A-duration of the impulse is ≈ 0.3 milliseconds instead of ≈ 2 milliseconds, the peak pressure attenuation of the RACAL Gunfender earplug is 5 dB larger and the peak pressure attenuation of the double protection is 20 dB larger. We can state that it is impossible to assign one single attenuation value (either for the peak pressure or for the L_{Aeq8}) to an HP, even when it is linear. Peak pressure and/or L_{Aeq8} attenuation values cannot be used to characterize the performance of HP in high-level impulse noise in a simple way. Finally, it should be noted that for a given HP there is no direct relation between the peak pressure under the HP (or the peak-to-peak pressure) and the NR (or the IL) amplitude spectrum measured in the same exposure conditions.

One can speculate about the interpretation of the attenuation values calculated from sig-

nals recorded in the free field and under an HP (NR values). It may appear more logical to consider using the peak pressure and the L_{Aeq8} attenuation values measured at the same location on protected and unprotected ears (IL values), preferably at the tympanum, to obtain comparable values for all kinds of HP. Actually, in the case of high-level impulses these measurements are feasible only on an ATF. Figure 26-13 presents the pressure-time signature of an impulse (170 dB peak, 2 milliseconds A-duration) measured simultaneously in the free field and at the ear simulator microphone for normal incidence in the unprotected condition, and protected with a RACAL Airsoft earplug. Because of the TFOE, there is an amplification of the peak pressure by 13 dB from the free field (170 dB) to the ear simulator microphone (183 dB) in the unprotected condition. The peak attenua-

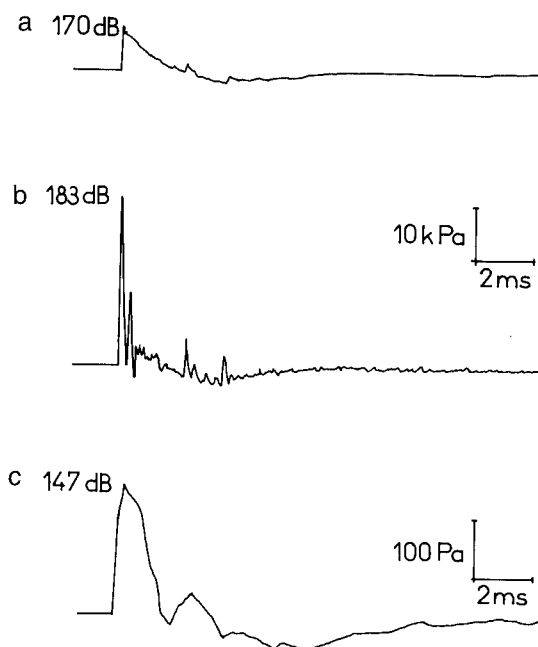


Figure 26-13 Pressure time signatures recorded (a) in the free field (peak pressure, 170 dB; A-duration, 2 milliseconds), (b) at the microphone of the ISL ATF ear simulator (unprotected, normal incidence) (peak pressure, 183 dB), and (c) at the microphone of the ISL ATF ear simulator (protected by a RACAL Airsoft earplug, normal incidence) (peak pressure, 147 dB).

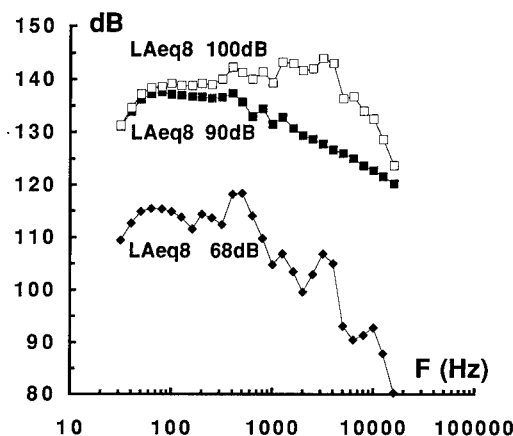


Figure 26-14 Amplitude spectra and L_{Aeq8} values (■) of the impulses in the free field, (□) of the impulses recorded by the unprotected ear simulator microphone (normal incidence), and (◆) of the impulse recorded by the protected (RACAL Airsoft earplug) ear simulator microphone ($1/3$ -octave bands).

tion from free field to the microphone in the presence of the HP is 23 dB (see Figure 26-13 and Table 26-1), whereas the peak attenuation measured by comparing the pressure-time signatures measured by the microphone on the unprotected and protected ear (IL value) is 36 dB. This last figure does correspond to the actual attenuation of the peak pressure received by the ear when equipped with the RACAL Airsoft earplug in our experimental conditions. If we look now at the amplitude spectra and the L_{Aeq8} of these signals (Figure 26-14), we notice that although the attenuation of the A-weighted acoustic energy (L_{Aeq8}) from the free field to the ear simulator microphone is 22 dB, the actual attenuation of the L_{Aeq8} afforded by the same earplug is 32 dB at the ear.

In conclusion the attenuation measurements calculated from signals recorded in the free field and under an HP are not representative of the actual protection and may grossly underestimate it. This is especially true in the case of earplugs (attenuation measurements performed under the cup of an earmuff would be less misleading). Hence, the only valid attenuation values (peak pressure, L_{Aeq8}) are those obtained from signals measured at the

same location (tympanum) on the protected and unprotected ear, that is, the attenuation values corresponding to the insertion loss.

Hearing Protection in High-Level Impulse Noise

In practice, in high-level impulse noise there are two main methods available to decide whether the hearing protection afforded by a HP is sufficient. The first is by measuring the signal close to the head of the subject and using the IL characteristics of the HP (NR characteristics are inadequate because they lead to an underevaluation of the actual protection, especially for earplugs) corresponding to the type of impulse (because of possible nonlinearities, the IL values obtained from REAT methods are not always representative of the actual attenuation afforded by HP for high-level impulses) to calculate the equivalent dose of acoustic energy to which the subject would be exposed unprotected. The second method is by measuring the pressure-time signature of the impulse under the HP and introducing the measured parameters (i.e., peak pressure and duration) into the classical DRC for weapon noises.¹

The second possibility^{8,9} involves an untested extension of the use of the weapon noise criteria because these various DRC¹ have been primarily designed to apply to pressure-time signatures measured in the free field and to unprotected ears. Moreover, as explained, the physical meaning of pressure measurements under an HP differs according to whether the microphone is placed under an earmuff (i.e., usually close to the entrance of the auditory canal) or under an earplug (i.e., close to the tympanum). In the first case the transfer function of the auditory canal is not taken into account; in the second case this transfer function (modified by the insertion of the earplug) plays a role (this difficulty would disappear if all measurements were performed close to the tympanum).

In some instances, a global protection factor corresponding to a typical HP has been applied to the peak pressure of the impulse be-

fore its evaluation with the help of a DRC.³³ This method is incorrect too because there is no direct relation between the peak pressure attenuation of an impulse afforded by an HP and the IL characteristics of the same HP (even for a linear HP).

The problem is which methods yields the most representative estimate of the actual hearing protection afforded by HP? To answer this question we have two sets of recent experimental results. Soldiers equipped with different HP were exposed to high-level impulse noises by Dancer et al.,³⁴ Johnson and Patterson,¹¹ and Patterson et al.,¹² and audiometric tests were performed just before and after the exposures. Dancer et al.³⁴ exposed 20 subjects wearing well-fitted E·A·R foam earplugs to 20 rounds of howitzer muzzle noise (peak pressure, ≈ 175 dB; A-duration, ≈ 8 milliseconds; global L_{Aeq8} , 109 dB). Only 1 ear out of 40 exhibited a temporary TS (TTS) larger than 10 dB (15 dB at 6 kHz). This TTS recovered 1 hour later. The peak pressure attenuation afforded by the E·A·R foam earplug, measured by means of the microphone of the ATF ear simulator in similar exposure conditions (by comparing the pressure time signatures in protected and unprotected conditions; IL measurements) is around 30 dB.¹⁵ If we enter this IL attenuation value into the classical DRC for weapon noises (see Figure 26-15 according to Smoorenburg³⁵) we observe that this exposure is just on the limit. On the other hand, the L_{Aeq8} attenuation corresponding to the IL measurements is about 30 dB. A subject equipped with the E·A·R foam earplugs, exposed to the 20 rounds of the howitzer (global L_{Aeq8} , 109 dB on the outside), is in the same exposure conditions as an unprotected subject exposed to a L_{Aeq8} of about 80 dB (as far as the A-weighting and isoenergy principles are valid for such exposures). Dancer et al.³⁴ exposed another 11 subjects wearing well-fitted RACAL Gunfender earplugs to 10 rounds of the same howitzer (peak pressure, ≈ 175 dB; A-duration, ≈ 8 milliseconds; global L_{Aeq8} , 106 dB). No TTS larger than 10 dB was observed at any frequency. The peak pressure attenuation and the L_{Aeq8} attenuation (IL measurements)

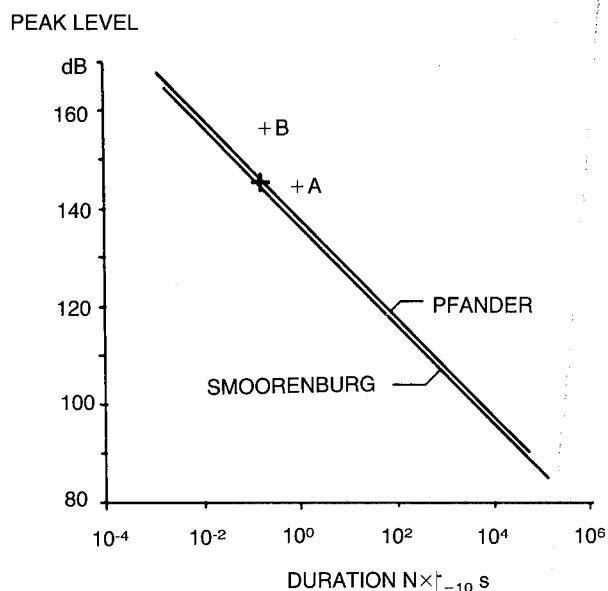


Figure 26-15 Damage risk criteria for weapon noises of Pfander and Smoorenburg.³⁴ The abscissa gives the total duration of all impulses, given by the number, N , of the impulses multiplied by the duration at -10 dB from the peak, of each impulse. (a) exposure conditions corresponding to 20 rounds of 175 dB peak pressure (duration at -10 dB, 25 milliseconds) with an E·A·R foam earplug and (b) exposure conditions corresponding to 10 rounds of 175 dB peak pressure (duration at -10 dB, 25 milliseconds) with a RACAL Gunfender earplug.

afforded by the RACAL Gunfender earplug are about 20 dB.¹⁵ If we enter the peak pressure attenuation (IL value) into the DRC for weapon noises, we observe that this exposure is well beyond the limits. On the other hand, a subject equipped with these earplugs and exposed to 10 rounds of the howitzer would have the same exposure conditions as an unprotected subject exposed to an L_{Aeq8} of about 86 dB. In these experiments, unlike the peak pressure attenuation values (even when based on IL measurements), the L_{Aeq8} attenuation values based upon IL measurements might explain the essential absence of auditory fatigue in the subjects exposed to these large impulses.

In another experiment conducted by Johnson and Patterson¹¹ and Patterson et al.,¹² 96 subjects were exposed to 100 impulses of 187 dB peak pressure (A-duration, 3 milliseconds at 1 minute intervals (global L_{Aeq8} , 133 dB)

wearing a well-fitted standard muff (SM). Virtually no TTS at any frequency was observed. In another experiment 57 subjects were exposed to the same impulses but wearing a modified muff mimicking a leaky one (MM). Only one subject presented TTS. The peak pressure attenuation from the free field to under the muff (NR measurements) was 12 dB for the SM and 8 dB for the MM. These attenuation values correspond to totally unacceptable exposures according to the classical DRC. The L_{Aeq8} attenuation from free field to under the muff (NR measurements) was 23 dB for the SM and 15 dB for the MM. Unfortunately, the attenuation values based on IL measurements are not available from the studies of Johnson and Patterson¹¹ or Patterson et al.¹² It is likely that these values are somewhat larger than those indicated above, nevertheless the L_{Aeq8} corresponding to the exposure of an unprotected subject would

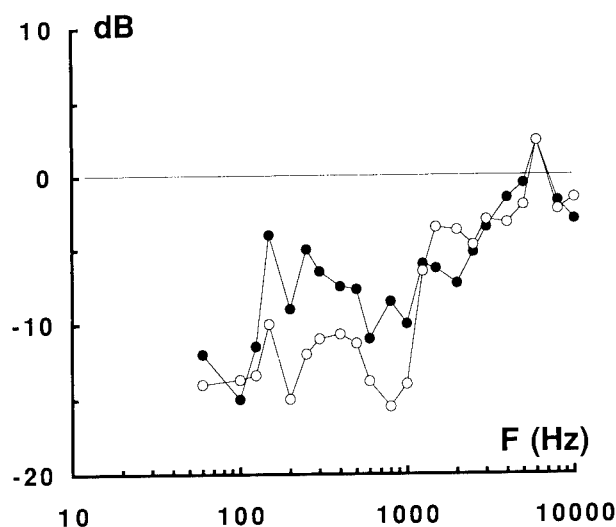


Figure 26-16 Modification of the amplitude of the middle ear transfer function in humans at the tympanum (mean results on 8 subjects) by application of DC pressures: (●) positive, 5 kPa; (○) negative, 5 kPa.

probably be beyond 100 dB for the SM and 108 dB for the MM. As pointed out by Johnson and Patterson¹¹: "... the use of peak level reduction as a measure of hearing protection underestimates the protection given. A-weighted energy is a step in the right direction, but its use still underestimated the performance of the hearing protectors used in our study."

From these NIHL studies we can conclude the following.

(a) It is actually possible to protect the ear when exposed to high- and very high-level impulse noises with the help of single standard HP (either plugs or muffs) despite the fact that the peak pressure at the protected ear is well over the 140 dB limit fixed by the standards (up to 165 dB in the Dancer et al.³⁴ experiment, and up to 180 dB in the Johnson and Patterson¹¹ experiment). These observations do question the use of the "peak pressure" as a relevant parameter to evaluate the hearing hazard on protected ears. The risk corresponding to the exposure to a slow rise time impulse (as recorded under an HP) is in fact much lower than the risk corresponding to the exposure to a Friedlander wave with an

almost instantaneous rise time (with the same peak pressure). A possible explanation of this phenomenon is in the influence of the spectral factor³⁶⁻³⁸ alone or in combination with the nonlinear mechanisms at the level of middle and inner ears.³⁹⁻⁴²

(b) The peak pressure attenuation (based on IL measurements and all the more so on NR measurements) grossly underestimates the protection afforded by the HP when used in conjunction with the DRC for weapon noises.¹

(c) The L_{Aeq8} attenuation (based on IL measurements) gives in some instances a good evaluation of the auditory hazard (as in the Dancer et al. experiments with plugs³⁴), but fails in some others to evaluate correctly the efficiency of the HP (Patterson et al. experiments with muffs¹²). Among the conventional methods used in evaluating hazards, the use of the L_{Aeq8} attenuation values based on IL measurements seems to represent the best conservative approximation.

Better approximations would require that at least the nonlinearity of the middle ear be taken into account. Price and Kalb⁴¹ empha-

sized the limitation of the tympano-ossicular chain displacements due to the nonlinear mechanical characteristics of the annular ligament when exposed to large impulses. If important for unprotected exposures, this effect could be essential in understanding the surprisingly small damage induced by large but slow-rising impulses such as those existing under HP. According to Price and Kalb,⁴¹ under such exposure conditions the nonlinearity of the middle ear transfer function could induce a protection of the sensory structures because the very large low-frequency components of the signals could block the transmission of the higher frequencies by the middle ear.

To assess the importance of this phenomenon, we measured the changes in the middle ear transfer function in eight human subjects during application of positive and negative DC pressures at the tympanum⁴³ (by recording the auditory sensitivity thresholds) (Figure 26-16). For a DC pressure of 5 kPa, the amplitude of the middle ear transfer function is lowered by about 10 dB up to 1 kHz and by 5 dB at 2 and 3 kHz. These results indicate a possible protective effect due to the low-frequency components on the midrange frequencies and could partially explain the results obtained by Johnson and Patterson¹¹ and Patterson et al.¹²

Furthermore, in high-level impulse noise the mechanisms producing damage at the cochlear level likely depend in a nonlinear way on the maximum displacement of the sensory structures (the isoenergy principle is probably no longer valid). A moderate increase of the mechanical stress may induce a large increase in cochlear sensory cell losses. This phenomenon is generally ascribed to the existence of a "critical level."⁴⁴⁻⁴⁷ Thus, a relatively small attenuation of the peak pressure of the impulses could help prevent instantaneous damage to the organ of Corti.

Conclusion

Many hearing protectors behave nonlinearly in high-level noise. This nonlinearity can be

favorable (increase of the attenuation with the level) or unfavorable depending on the type of HP and on the pressure-time signature of the sound waves.

The attenuation characteristics (noise reduction and insertion loss values, etc.) of HP that are intended to be used in high-level acoustic environments must be measured under the same exposure conditions.

The easiest and most reliable way to assess the physical attenuation provided by HP (either earmuffs or earplugs) in any exposure condition is to use an ATF equipped with an ear simulator, and to perform the measurements at the level of the "tympanum" (microphone of the ear simulator). The ATF must present a dynamic range large enough to allow the evaluation of HP at intensities as high as those encountered in the actual situation.

Presently the best estimate of the protection afforded by an HP seems to be given by the attenuation value of the A-weighted acoustic energy (i.e., attenuation of the L_{Aeq8}) measured at the level of the "tympanum" (IL value) when the HP is worn.

However, the HP is not the only nonlinear element that needs to be taken into account when studying the NIHL hazards on protected ears in high-level (impulse) noise. According to Price and Kalb,⁴¹ the middle ear and the cochlea behave nonlinearly in extreme environments. To build a more comprehensive DRC for weapon noises, we must take into account the nonlinear behavior of the HP as well as that of the ear and to study their interrelations. To this end, recording of the pressure-time signatures in free field and at the tympanum on unprotected and protected ears will be of the utmost importance. In most opportunities these recordings will be possible only with the help of a specially designed ATF.

Acknowledgment

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Chapter 27

Assessment of Hearing Protector Performance in Impulsive Noise: Update of Research Activities Within the EC-Funded IMPRO Project

Adelbert W. Bronkhorst and Guido F. Smoorenburg

ISO 4896-1 is a standardized method for assessment of hearing protector attenuation.¹ This method is based on subjective threshold measurements for narrowband noise stimuli at several center frequencies. The attenuation is taken as the difference between thresholds with and without hearing protectors. Advantages of this method are that it is based on sounds that actually reach the inner ear (including bone-conducted sounds), and that it includes the effect of intersubject differences in hearing protector fit. However, it has three important disadvantages:

1. Hearing protector attenuation is determined only for low sound levels. Thus, performance of level-dependent hearing protectors, such as sound-restoring devices and devices with active noise reduction, cannot be assessed.
2. The method yields attenuation values at only 7 or 8 octave frequencies and provides no information on phase. It is therefore questionable whether the response of hearing protectors to impulse noise (e.g., the peak level attenuation) can be reliably estimated from the results.
3. Hearing protector attenuation below 500 Hz is overestimated due to the effect of physiological noise. This has been revealed by sound level measurements in the human ear canal reported in several studies.^{2,3}

In view of the increasing use of nonlinear hearing protectors, and given that both maximum allowable peak levels and hearing protector attenuation are subject to European Community (EC) directives, it is mandatory that new methods for assessment of hearing protector performance are developed. The aim of the IMPRO (impulse protection) project, funded by BCR (Bureau Communautaire de Référence) and carried out by a consortium of nine research groups, is to investigate whether acoustic measurements can be used for this purpose.* As indicated by its name, the IMPRO project is primarily aimed at assessment of hearing protector performance in impulsive noise. The project was preceded by two feasibility studies. The first study, carried out at the University of Salford in the United Kingdom, identified impulsive noise sources that can be used to test hearing protectors in the laboratory.⁴ The second study, conducted

*Participating laboratories within the EC are: (1) The Acoustics Laboratory, Technical University of Denmark (ALTUD, DK); (2) Berufsgenossenschaftliches Institut für Arbeitssicherheit (BIA, D); (3) Institut National de Recherche et de Sécurité (INRS, F); (4) Institut Franco-Allemand de Recherches de Saint Louis (ISL, F); (5) Physikalisch-Technische Bundesanstalt (PTB, D); (6) TNO Human Factors Research Institute (TNO, NL); (7) University of Salford, Department of Applied Acoustics (USDAA, UK). Participants outside the EC are (8) Karolinska Institutet, Unit of Technical Audiology (KITA, S) and (9) The Finnish Institute of Occupational Health (FIOH, SF).

by TNO (Organisatie voor Toegepast Natuurwetenschappelijk Onderzoek) in the Netherlands, investigated the feasibility of probe microphone measurements in the human ear canal (the MIRE, microphone in real ear, technique) for determining sound attenuation of earmuffs.³ The IMPRO project started in the beginning of 1993 and will terminate in 1995.

Research Within IMPRO Project

The research within the IMPRO project encompasses a broad range of activities: new measuring techniques are developed, extensive measurements are conducted both in the laboratory and in actual workplaces, a literature survey is carried out, and central analysis and evaluation of the obtained data is performed. The activities fall into the following four categories:

(1) A MIRE technique for measuring sounds under ear plugs, suitable for all types of plugs including custom-molded ones, is developed. Requirements are that the technique does not affect the performance of the plug, that it can be performed quickly and simply, and that it presents no danger to the subject.

(2) MIRE measurements and simultaneous measurements outside the hearing protector (muff or plug) of steady-state and impulsive sounds are made. The measurements are performed both in the laboratory and in actual workplaces. They can be used to trace nonlinearities in hearing protector performance and to determine attenuation of peak levels of impulsive sounds. Attention is also given to the dependence of hearing protector attenuation on angle of incidence.

(3) Measurements are made of hearing protector attenuation for steady-state and impulsive sounds performed with ATFs (acoustic test fixtures). Advantages of ATFs are that the obtained results are reproducible and that measurements can be performed at high sound levels, because no human ear is involved. Disadvantages are that sound leakage may occur, especially in artificial heads that

are not designed for these measurements, and that no representative hearing protector fit may be obtained, in particular with earplugs.

(4) Analysis and evaluation of the data are performed. Results including DAT tapes with sound recordings are collected and processed at one location to insure compatibility, integration, and comprehensive reporting. The literature on the relationship between measures of impulse sound and risk of hearing loss is evaluated. The aim is to obtain a simple but adequate method of quantifying the attenuation of linear and level-dependent hearing protectors in impulsive noise with regard to risk of hearing loss.

In the two following chapters, a number of results obtained during the first 16 months of the project will be highlighted. Attention will be given to the MIRE technique applied to earplugs, and to the analysis of the simultaneous sound recordings under and outside hearing protectors. MIRE measurements with plugs are of interest because it turns out to be very difficult, perhaps even impossible, to satisfy demands of accuracy, feasibility, and subject safety while leaving the plug intact. The sound recordings provide detailed insight into the time and level-dependent response of hearing protectors to impulsive sounds.

MIRE Technique for Earplugs

By performing acoustic measurements in the human ear canal, hearing protector performance can be determined in a relatively straightforward manner. The attenuation as a function of frequency can be readily obtained by subtracting frequency spectra of a broadband sound measured with and without a protector.[†] Measurements in the ear canal have been greatly facilitated by the development, in the last decades, of small, high quality electric microphones. These microphones can either be equipped with a probe tube ex-

[†]However, when the attenuation of the hearing protector approaches the bone-conduction threshold, a correction should be applied to the results of the acoustic measurements.⁵

tending into the ear canal or the microphone itself can be inserted into the ear canal.

To measure the sound that is actually entering the ear, the microphone (or probe tube) should, in principle, be placed close to the ear drum. This is, however, not feasible for routine testing because such a placement is difficult to obtain and possibly hazardous to the test subject. For earmuffs, the choice of microphone/probe tube location was investigated in one of the feasibility studies preceding the IMPRO project. Attenuation values measured with a miniature microphone in the concha, either without a probe tube or equipped with a 12.5 or 25 mm probe tube, were compared with results of subjective threshold measurements (REAT, real ear attenuation at threshold). The closest correspondence between MIRE and REAT results was found for a probe tube length of 12.5 mm, that is, when the entrance of the probe is approximately halfway into the ear canal. This probe tube length is used in all MIRE measurements under earmuffs performed within the IMPRO project. The microphone is connected to a frame that can be attached in a simple manner to the outer ear.

Whereas development and verification of the MIRE technique for earmuffs proved to be relatively straightforward, application of the technique to earplugs presents considerable difficulties. In the experiments described, two different approaches are used. In the first approach, a subminiature microphone is placed under the earplug. Although this has the advantage of leaving the plug intact, it appears that a good fixation of the microphone and a reliable electrical connection along or through the plug are not easily achieved. In addition, the measurement is not without danger for the test subject, as the microphone may touch or even damage the ear canal. In some cases, it may even be impossible to fit the microphone into the ear canal. In the second approach, the microphone is placed outside the plug, using a probe tube or a hole bored through the plug for the sound measurement. Disadvantages of this method are that sound leakage may occur when a probe tube is passed along the plug, and that boring a canal

through the plug may change the properties of the plug itself. Only when a canal is already present (as is the case in certain custom-molded types) will the measurements not affect the performance of the plug. A third approach, until now only tested in preliminary measurements, is placement of the microphone inside the plug. Naturally, this method may also affect the properties of the plug.

Results of an experiment with custom-molded plugs, performed at the INRS laboratory in France, are presented in Figure 27-1. Shown is the plug attenuation, averaged over eight subjects and measured in three different ways, as a function of octave frequency. The REAT results, indicated by the diamonds, were obtained using the measuring procedure described in ISO 4869-1.¹ For the MIRE measurements, performed with a Knowles EM 3046 subminiature microphone, a hole was bored through the plug. The open squares show results that were obtained with the microphone outside and a probe tube inserted through the hole. In this case, the unoccluded measurement was performed with the same

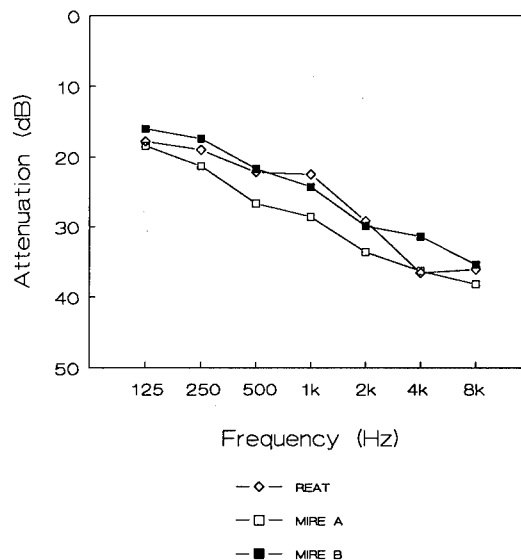


Figure 27-1 Attenuation of custom-molded earplugs as a function of frequency, measured in three ways: subjectively (REAT), with a microphone outside and probe tube through the plug (MIRE A), and with a microphone under the plug (MIRE B). Average results for eight subjects.

microphone and probe tube placed in the open ear. The closed squares indicate results obtained with the microphone under the plug. The connecting wires were passed through the hole that was sealed with mastic. For the open-ear measurement, the microphone was placed at approximately the same position in the ear canal. In both MIRE measurements, earplug attenuation was taken as the difference between $\frac{1}{3}$ -octave noise levels measured in the open and occluded ear. The figure shows that MIRE results obtained with a microphone under the plug are in close agreement with the REAT data. The other MIRE technique appears to overestimate actual plug attenuation.

At PTB in Germany, measurements with foam plugs were performed. Because it proved to be very difficult to place a microphone behind the plug, the insert earphone ER 3A, similar to the E-A-R foam plug, was used in the experiment. For the MIRE measurements, the microphone (a Sennheiser KE 4-211-2) was connected to the tube passing through the plug. During the REAT measurement, the tube was closed with a small screw. The open ear MIRE measurement was performed with a tube from which the foam plug was removed. Figure 27-2 shows average results for 16 subjects. REAT and MIRE results are indicated by the diamonds and the squares, respectively. For frequencies above 500 Hz, there is an excellent agreement between the results of both methods. At lower frequencies, the REAT method yields higher attenuation values than the MIRE method. The difference is too large to be contributed solely to the effect of physiological noise. It may also be due to the fact that the REAT measurement was not performed with the microphone in place.

At TNO, experiments with custom-molded earplugs were carried out. The plugs were equipped with an acoustic filter and already had a canal bored through them. First, a sub-miniature microphone (the Knowles EM 3046) was mounted on the inside of the plug, using the canal for the electrical wiring while leaving the filter intact. However, a good fit of the plug could not be achieved with this method because the microphone touched the ear canal. This also caused discomfort to the test

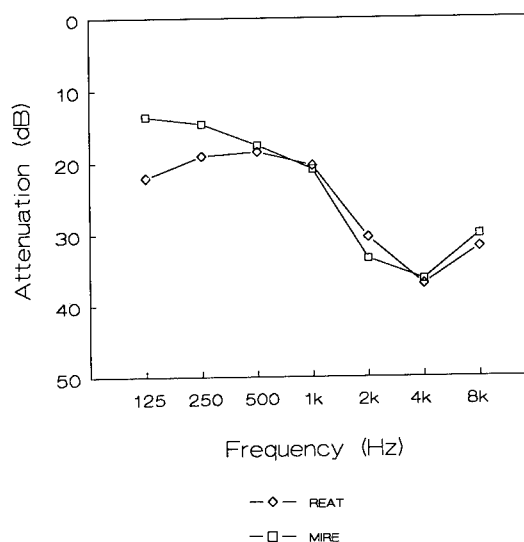


Figure 27-2 Subjective (REAT) and objective (MIRE) measurements of the attenuation of foam plugs as a function of frequency. The microphone was placed outside the plug. Results are averaged over 16 subjects.

subjects. It was therefore decided to place the microphone outside the plug. The acoustic filter was replaced by a hollow plastic plug containing a Sennheiser KE-211-9 microphone and an extra hole for the filter. Thus, the canal could be used for the sound measurement while maintaining the effect of the filter. The experimental results, obtained with six subjects, are shown in Figure 27-3. REAT measurements were performed both with the original earplug (the open diamonds in the figure) and with the plug that had the microphone attached to it (the closed diamonds). The plug was replaced between these measurements. The MIRE measurement was performed directly after the second REAT measurement. For the open ear measurement, a probe tube, having approximately the same length as the canal through the plug, was attached to the microphone plug and inserted into the ear canal. This measurement was corrected for the difference in frequency response between the probe tube and the canal in the earplug. The MIRE results, indicated by the squares, were obtained by subtracting the $\frac{1}{3}$ -octave noise levels measured in the occluded ear

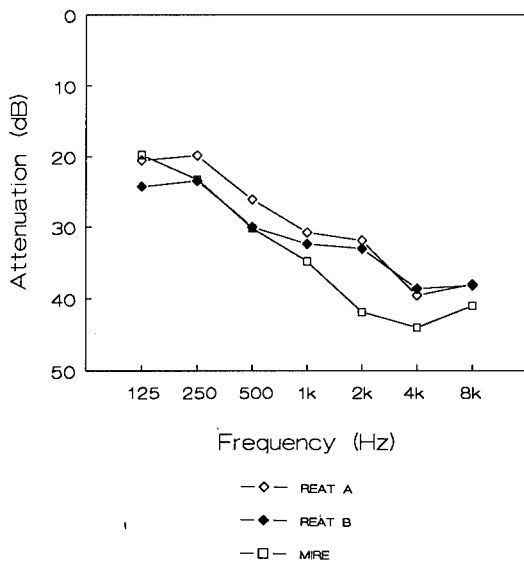


Figure 27-3 Attenuation of custom-molded earplugs as a function of frequency, measured in three ways: subjectively with an intact plug (REAT A), subjectively with a plug that has a microphone enclosure attached to it (REAT B), and objectively using the microphone (MIRE). Average results for six subjects.

from the (corrected) levels obtained in the unoccluded ear. The figure shows that the results of the two REAT measurements are almost the same, meaning that placement of the plug with the microphone does not affect the plug attenuation. The MIRE results, however, show a deviation from the REAT results, especially at 2 and 4 kHz. This is probably due to resonances in the probe tube and in the canal through the earplug that are insufficiently compensated for by the correction applied to the open ear measurements.

Simultaneous Sound Measurements Outside and Under Hearing Protectors

When sounds outside and under the hearing protector are measured simultaneously, the behavior of the protector in different types of noise can be analyzed in detail. The complex-valued transfer function of the protector, that is, not only its attenuation but also the phase response, can be obtained.⁶ This transfer func-

tion can be determined for various sound levels and as a function of the time after onset of an impulse to trace nonlinearities in the response of the protector. Given that the protector behaves linearly, the waveform under the protector, including the peak level, can be calculated for any given sound occurring outside.

During all field measurements conducted within the IMPRO project, simultaneous sound recordings outside and inside the hearing protector were performed using the MIRE technique and a sound-level meter placed on the shoulder of the test subject. Preliminary recordings, obtained by several research groups, were analyzed at TNO. A software tool was used that performs real-time calculation of the transfer function and real-time linear filtering. The software runs on a PC equipped with a DSP-board.

Table 27-1 shows measured and calculated peak levels in decibels sound pressure level (dB SPL) for two muffs and three signal types. The sound recordings took place at BIA in Germany (drum beats) and at ALTUD in Denmark (pistol shots and strokes of a hammer on a plate). They were performed with a B&K sound-level meter, placed on or above the shoulder, and a Sennheiser KE-211-9 with probe tube, placed in the ear. Each combination of muff and signal was measured for two subjects. The results for the drum beats are averaged over eight impulses; those for the shots and hammer strokes over three impulses. Both over- and underpressure peak levels were determined. The maximum of the two was entered into the table. Transfer functions of the muffs and of the open ear were determined from recordings of steady-state broadband noise presented in a free field. The calculated peak levels in the occluded and open ear were obtained by filtering the signal, recorded outside, according to the transfer function of the muff and open ear, respectively. The Δ column shows the differences between measured and calculated peak levels in the ear. The last column shows the difference between the (calculated) open and (measured) occluded ear peak levels, that is, the peak attenuation of the muff.

Table 27-1 Measured and Estimated Peak Levels (dB SPL) and Peak Attenuations (dB) for Two Muffs and Three Signals

Muff Type	Subject	Signal	Measured		Calculated			Attenuation
			Outside	In Ear	In Ear	Δ	Open Ear	
No muff	1	Drum beat	132.7	139.5		-2.3	137.2	
	2		134.4	140.4		0.7	141.1	
Peltor H7A	1	Drum beat	132.9	114.8	115.1	0.3	138.6	23.8
	2		135.6	110.9	113.3	2.4	141.7	30.8
	3	Pistol shot	148.8	119.5	116.6	-2.9	156.7	37.2
	4		141.8	123.7	124.1	0.4	160.3	36.6
	3	Hammer on plate	143.3	113.9	114.9	1.0	161.0	47.1
E·A·R Ultra 9000	4		137.1	113.8	121.4	7.6	157.4	43.6
	1	Drum beat	131.5	126.2	122.7	-3.5	137.6	11.4
	2		136.3	119.2	122.3	3.1	144.0	24.8
	3	Pistol shot	146.6	134.6	133.9	-0.7	156.6	22.0
	4		140.8	130.8	135.0	4.2	157.7	26.9
	3	Hammer on plate	139.5	131.0	136.6	5.6	160.8	29.8
	4		135.5	127.6	133.5	5.9	157.3	29.7

It appears that the peak levels are not predicted exactly by the linear filtering. In particular for the hammer strokes, the deviation can be quite large. There are several possible causes of these discrepancies. First, in several cases, the positions of the microphones, the orientation of the sound source, and the fit of the hearing protector were not the same during the noise recordings and the impulse recordings. Second, the acoustics of the recording environment was sometimes changed. This particularly has an effect with long-duration impulses, like the hammer strokes. Third, the hearing protector might show nonlinear effects. This, in fact, should be the case with the E·A·R Ultra 9000, which is designed to provide higher attenuation with increasing peak levels. The calculated peak levels do not show such an effect, because they are not significantly larger than the measured levels. However, the nonlinear effect is revealed in another analysis technique, discussed below.

It should be stressed that the difference between the peak levels outside and under the hearing protector does not provide a meaningful indication of the peak attenuation provided by the hearing protector, because the

acoustic effects of the ear canal and probe tube are not taken into account. A better indication is obtained when the difference is taken between the occluded ear and the open ear peak levels. In most cases, it is not possible to actually measure the open ear levels, but they can be estimated by linear filtering. The calculated open ear peak levels and the resulting peak attenuation are shown in Table 27-1 in the last two columns. It appears that the attenuation depends not only on the signal type, which is to be expected because of differences in frequency content, but also across subjects, especially for the drum beats. The E·A·R Ultra 9000 muff provides, on the average, 12 dB less peak attenuation than the Peltor muff.

The time-dependent behavior of the muffs was studied by calculating the transfer function of the muff for the pistol shots in short time frames of 20 milliseconds. The recordings were made at a distance of 1 m in a somewhat reverberant workshop. Results are shown in Figures 27-4 and 27-5. The attenuation of the muff in the first three frames, the first containing the shot, is plotted relative to the attenuation in the decaying sound field, measured during frames 5-14. Positive values indicate a

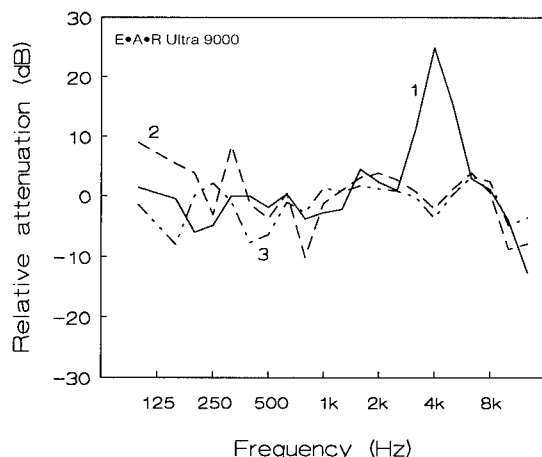


Figure 27-4 Attenuation in $1/3$ -octave bands of an E•A•R Ultra 9000 muff during and immediately following a pistol shot, plotted relative to its attenuation in the decaying sound field. Average results for three shots.

higher attenuation during the single frames. The results are averaged over three shots. The results for the E•A•R Ultra 9000, plotted in Figure 27-4, show a clear nonlinearity: the attenuation during the first frame raises by 25 dB at 4000 Hz. As shown in Figure 27-5, the nonlinearities in the response of the Peltor H7A remain within 10 dB.

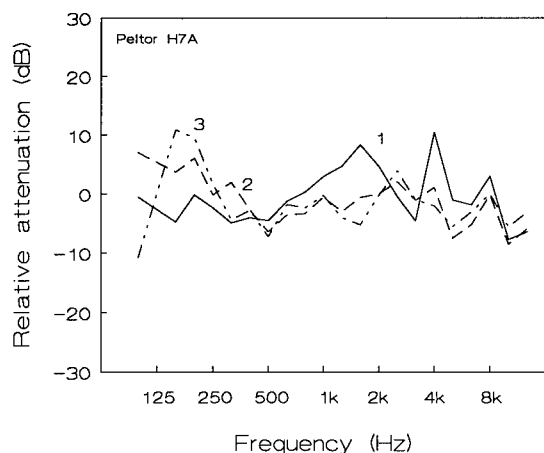


Figure 27-5 Similar data as shown in Figure 27-4, but obtained with a Peltor H7A muff.

Conclusion

A selection of results obtained within the EC-funded IMPRO project was presented. Experiments were performed to verify whether the MIRE technique, that is, the use of acoustic measurements in the ear canal, can be applied to earplugs. Results show that it is, in principle, possible to obtain attenuation estimates that are in close agreement with results of subjective threshold measurements. However, placement of the microphone at the inside of the plug, which is the best approach from a technical point of view, appears to be critical and sometimes even impossible to realize. As a consequence, it was decided to use only microphones that are either placed outside the plug or mounted inside of the plug in further experiments. To check whether this changes the properties of the plug, REAT measurements should be performed both with the original plug and with the plug adapted for the MIRE measurement. It is not yet clear whether the experiments will provide a method suitable for all types of plugs, that yields accurate results without affecting plug performance.

As part of the IMPRO project, extensive field measurements at workplaces and military locations will be carried out. They will consist of sound recordings performed simultaneously under and outside the hearing protector, using the MIRE technique and a sound-level meter placed on the shoulder of the test subject. The purpose is to gain insight into the time- and level-dependent behavior of hearing protectors. The recordings allow peak levels of impulse noises to be measured simultaneously outside and under the hearing protector. By applying linear filters to the recorded waveforms, the open ear peak levels can be predicted as well. Thus, the peak attenuation of the muff can be obtained, which is an important measure especially for nonlinear devices. In addition, the equivalent outside peak level can be calculated by applying an inverse filter to the waveform recorded in the occluded ear. This is relevant for application of directives that are formulated in

terms of free-field peak levels. The method of predicting peak levels by linear filtering has been applied to a number of preliminary recordings. Peak levels in the occluded ear were calculated and compared with measured levels. There appeared to be a reasonable agreement between both sets of results. However, for certain impulse noises relatively large differences were found. This indicates that application of linear filtering for prediction of peak levels requires further study. The two-channel recordings were also analyzed in the frequency domain. It appears that transient nonlinearities of hearing protectors can be revealed by a cross-spectrum analysis of the signals in short time frames.

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Chapter 28

Estimated Reductions in Noise-Induced Hearing Loss by Application of ANR Headsets

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A very rapid expansion of commercial headset manufacturing is currently underway. Since 1989, when the first commercially available active noise reduction (ANR) headset appeared on the commercial market in the United States, almost every major headset manufacturer has developed and marketed an ANR headset. Today, an estimated 12 or more different companies market a commercial ANR headset. Many of the early developments and evaluations of ANR headsets were accomplished by the military in Great Britain and the United States. The military, because of the high noise levels associated with military equipment, required noise attenuation beyond the available passive earmuff hearing protection devices. This ANR technology has now been employed in providing this increased noise attenuation in headsets that are commercially available from a large number of manufacturers.

ANR headset technology is significantly more expensive than passive headset technology. This is partly due to lack of competition in the early ANR headset market and partly due to the complexity of the ANR headset electronics. To determine which applications of ANR technology are satisfactory, several questions need to be answered. How much additional noise attenuation performance does the ANR headset provide beyond a good passive headset? Is the performance of all ANR headsets similar? What effect does the additional low-frequency attenuation of ANR headsets have on overall noise level at the ear? What is

the predicted reduction in noise-induced hearing loss and associated hearing loss compensation due to the application of ANR headsets when compared to passive headsets? Once these questions are answered for a given application, an informed selection can be made of a passive headset or an ANR headset.

Background

The concept of ANR was originally patented by a German scientist, Paul Leug, in 1936. ANR (or cancellation) reduces the overall level of noise by employing the technique of wave addition or wave cancellation (destructive interference). A miniature microphone placed inside the ear cup (of most ANR headset designs) measures the noise field inside the ear cup and sends a copy of the noise to an electronic circuit. The ANR system electronics invert the noise signal and send it back to the earphone inside the ear cup. If there were no time delays in the electronics and the earphone and microphone were perfectly matched and were in the same physical location, the noise signal from the earphone would be exactly 180° out of phase with the original noise signal and the cancellation would be perfect. However, the microphone and earphone cannot occupy the same location in space. Also, all the elements of the system add delays that limit the performance of the ANR system both in maximum frequency of active attenuation and in the achieved levels of active attenuation.

Development of practical ANR headset technology dates back to 1957 when Willard Meeker, under contract to our laboratory, demonstrated the feasibility of the ANR headset. Meeker described the basic control equations that are still used in almost all the currently available commercial ANR headsets. The ANR headset demonstrated by Meeker provided the basis for future ANR headset developments. P.D. Wheeler at the University of Southampton, under contract to Graham Rood, Royal Aircraft Establishment, United Kingdom, developed a flyable ANR headset system in the late 1970s. This system was the first ANR system to be demonstrated in actual flight conditions. The Wheeler ANR system had individual gain controls for the ANR electronics for each ear. The user would put on the headset and adjust the gain until the system went unstable and then reduce the gain until stability was satisfactory. In 1980, the United States Air Force, under contract with the Bose Corporation, initiated development of an ANR headset. Over a period of approximately 6 years, this effort resulted in a headset that did not require the user to adjust the ANR system and was stable under all normal use conditions. This system was demonstrated to provide improved attenuation, improved speech intelligibility in noise, improved comfort, and reduced fatigue.¹⁻³

In 1989, Bose introduced the first ANR headset that was available in large numbers to the general population. Since that introduction, the number of developers, manufacturers, and suppliers of ANR headsets has risen dramatically, as has the demand for ANR headsets.

Objective

The prices and the performance of ANR headsets encompass a very broad range. The objective of this effort was to estimate the reductions in predicted noise-induced hearing loss by application of ANR headsets in selected noise environments. This information is presented for all commercially available (in early 1994, through common sources) ANR headsets (within the delivery time limits of this

study) and for selected passive headsets. The information is intended to provide objective performance data and the estimated resultant effects on predicted noise-induced hearing loss. In addition, the methodology presented can be used in analyses of the efficacy of ANR applications in other noise environments.

Approach

The approach was to measure the laboratory attenuation of ANR and passive headsets. These attenuation performance values were then applied to selected noise environments and the A-weighted noise levels at the ear were calculated. The A-weighted noise values were then used to predict noise-induced hearing loss per ISO 1999 for both the ANR and passive headsets. Predicted reductions in noise-induced hearing loss and hearing loss compensation for application of ANR headsets in the selected noise environments were estimated using the calculated A-weighted noise levels at the ear. These performance data were then applied to real noise environments such as those found in multiengine propeller aircraft.

Method

Each of the six ANR and three passive headsets used in this study was purchased using normal commercial practices. The suppliers had no knowledge that the items were being purchased for this study. Therefore, the test items are representative only as a single example of a distribution of performances of the particular headset. The nine headsets used in this study are:

ANR headsets:

1. NCT NB-DX ANR,
2. David Clark DCNC-ANR,
3. Telex ANR 4000,
4. Bose Aviation,
5. Peltor ANR 7004,
6. Sennheiser HMC 200 ANR;

Passive headsets:

7. Sigtronics S-20,
8. Telex 2000E,
9. David Clark H10-60.

Attenuation was measured using the miniature microphone in real ear (MIRE) method. The MIRE data were measured with three repeat trials using 10 subjects, 5 female and 5 male giving a total of 30 open and 30 occluded measurements for each headset. Each subject was fitted with a Knowles 1834 miniature microphone positioned at the entrance to the ear canal for the sound field measurements. The microphone was attached to the center of a yellow foam earplug that occluded the ear canal and provided an average of 20 dB attenuation to the subject. The sound field was pink noise at an overall level of 105 and 115 dB. All of the $\frac{1}{3}$ -octave band sound field measurements were made using a Bruel & Kjaer 2131 real-time analyzer.

The unoccluded or open ear condition was measured and then the hearing protector was fitted by the subject. The experimenter visually inspected the hearing protector placement on the subject's head, and when the fit was satisfactory, the occluded condition data were collected. When the headset was an ANR headset, the subject activated the ANR system and a second measurement was made in the occluded condition. The difference between the open condition and the first occluded condition was calculated as the passive attenuation. The difference between the open condition and the second occluded condition when the ANR was in the active mode was calculated as the total attenuation for the ANR headsets. The active cancellation (only for the ANR headsets) was calculated as the difference between the total and passive attenuation at each of the $\frac{1}{3}$ -octave band test signals. Data were collected and averaged for both the left and right ears and for the attenuation. The entire procedure was conducted at both the 105 and 115 dB levels of ambient noise.

The A-weighted levels of noise at the ear were calculated using the following procedure. Mean attenuation data in $\frac{1}{3}$ -octave bands were calculated from the 30 trials for each device at the 105 dB sound field condition. The selected aircraft noise environments were measured in $\frac{1}{3}$ -octave bands. The A-weighted noise levels at the ear were computed by subtracting the attenuation values

from the noise, in $\frac{1}{3}$ -octave bands, and then applying the A-weighted values. The $\frac{1}{3}$ -octave band levels were then logarithmically summed to give the overall A-weighted level at the average ear.

Noise exposure criteria for hearing are based on the A-weighted sound pressure levels (SPLs) at the ears and the durations of exposures. Estimated noise-induced hearing loss was estimated from the calculated A-weighted noise levels and ISO 1999 using the assumptions of 20 years of job-related noise exposure.

Data

Figures 28-1–28-6 present total, passive, and active attenuation for the 105 dB SPL pink noise condition for the six ANR headsets purchased for this study. Figure 28-1 depicts the NCT NB-DX ANR attenuation performance. The NCT was the only headset in the study that did not employ a circumaural ear cup. The lack of an ear cup resulted in almost no passive attenuation up to about 4000 Hz. The active attenuation peaked at 200 Hz at 11 dB. Figure 28-2 shows the David Clark ANR headset performance at 105 dB. The passive attenuation shows no significant peaks or valleys and the active attenuation has a broad response from 8 dB at 31 Hz to 9 dB at 400 Hz. The 0 dB crossover was between 630 and 800 Hz. The total attenuation of the Telex ANR 4000 headset is shown in Figure 28-3. The passive attenuation of the Telex ANR is essentially 0 dB until 160 Hz, with a rapid increase (approximately 12 dB/octave) to a plateau of 13–16 dB from 315 to 630 Hz. The active attenuation is fairly broad with the 0 dB crossover occurring at 500 Hz and a maximum active attenuation of 9 dB. Figure 28-4 shows the performance of the Bose Aviation ANR headset. The passive attenuation is essentially 0 dB until 160 Hz, rising to 38 dB at 8 kHz. The active attenuation begins at 7 dB at 31 Hz, rising to 21 dB at 160 and 200 Hz and falling to 0 dB between 630 and 800 Hz. The Peltor ANR 7004 headset performance is shown in Figure 28-5. The passive attenuation begins at 5 dB at 31 Hz and is 11 dB by 160 Hz. The

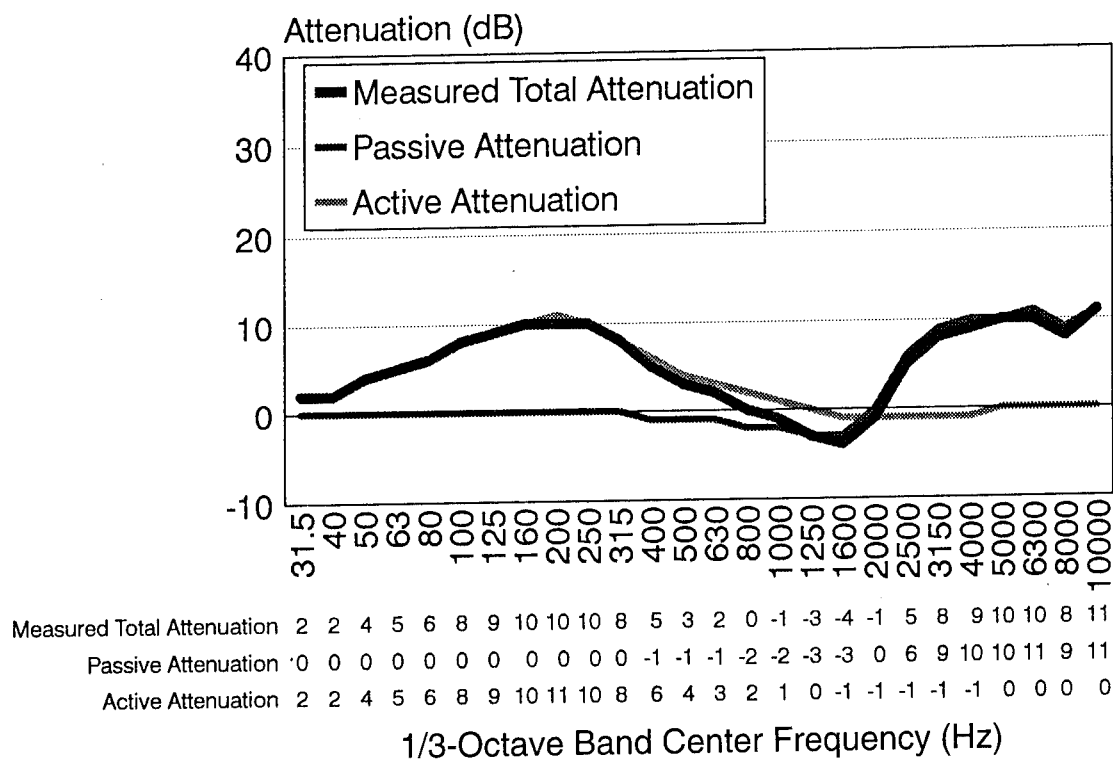


Figure 28-1 Active, passive, and total attenuation for the NCT NB-DX ANR headset.

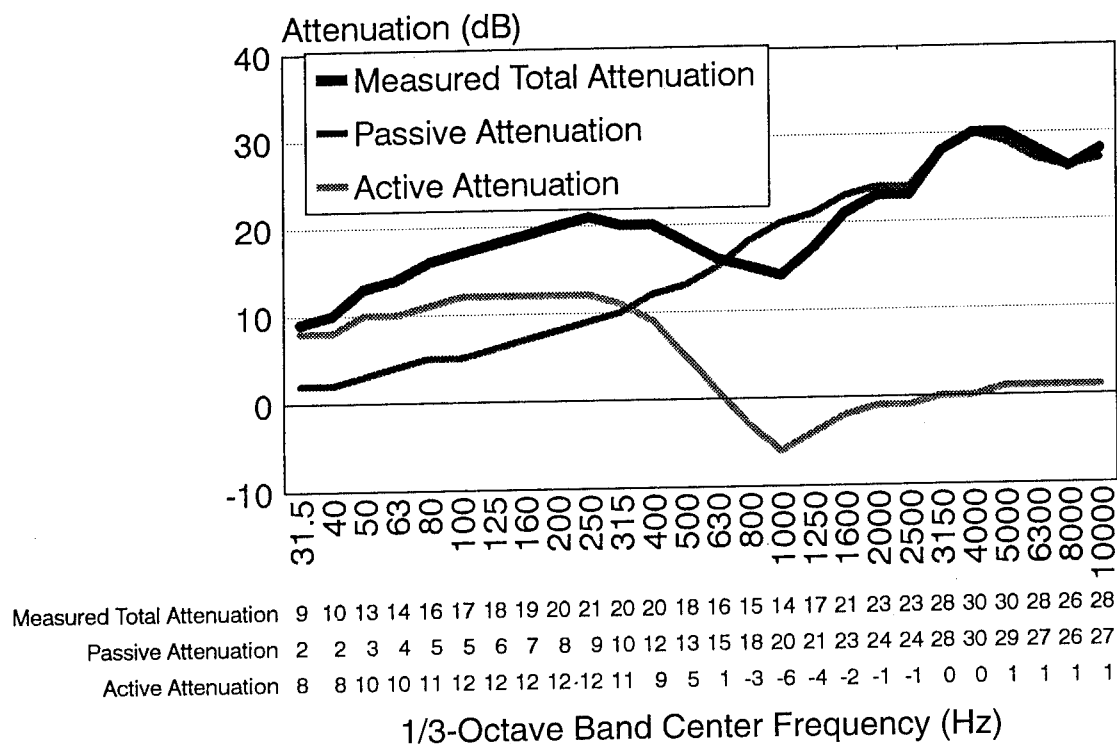


Figure 28-2 Active, passive, and total attenuation for the David Clark ANR headset.

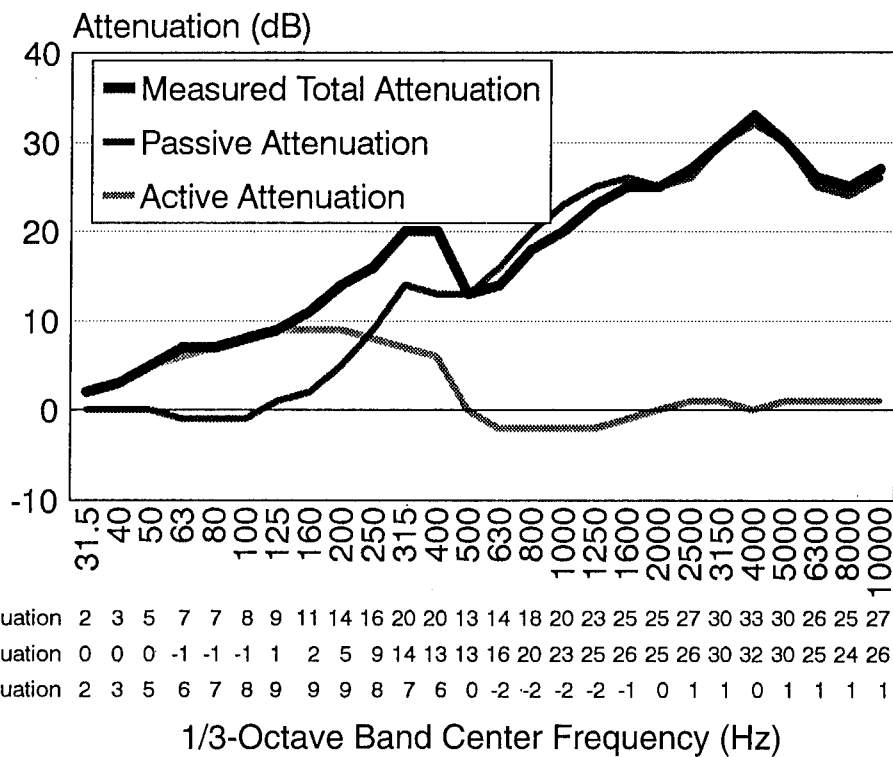


Figure 28-3 Active, passive, and total attenuation for the Telex ANR 4000 headset.

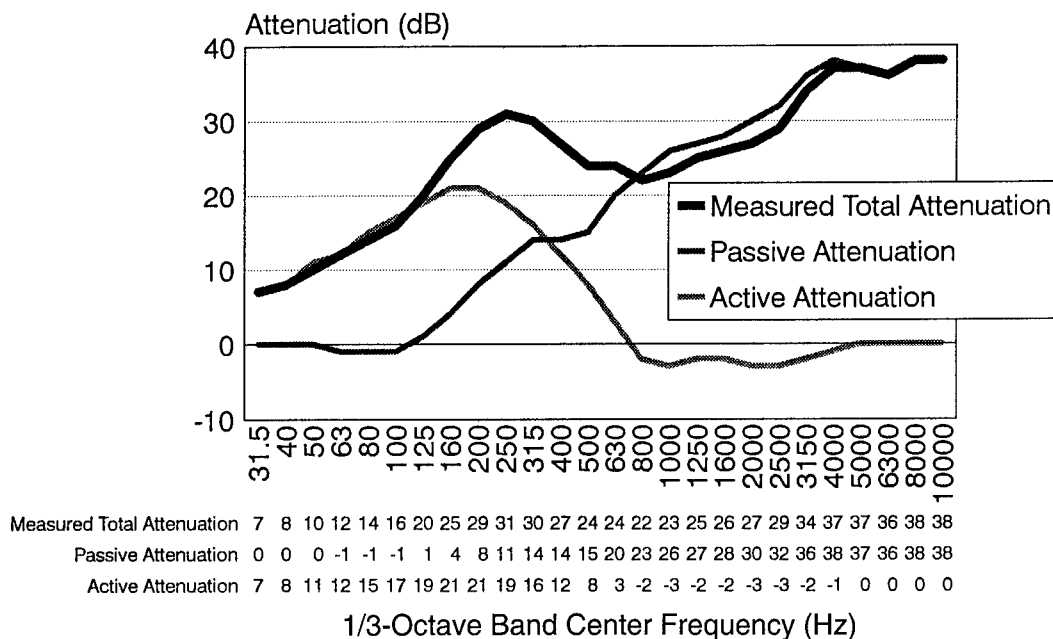


Figure 28-4 Active, passive, and total attenuation for the Bose Aviation headset.

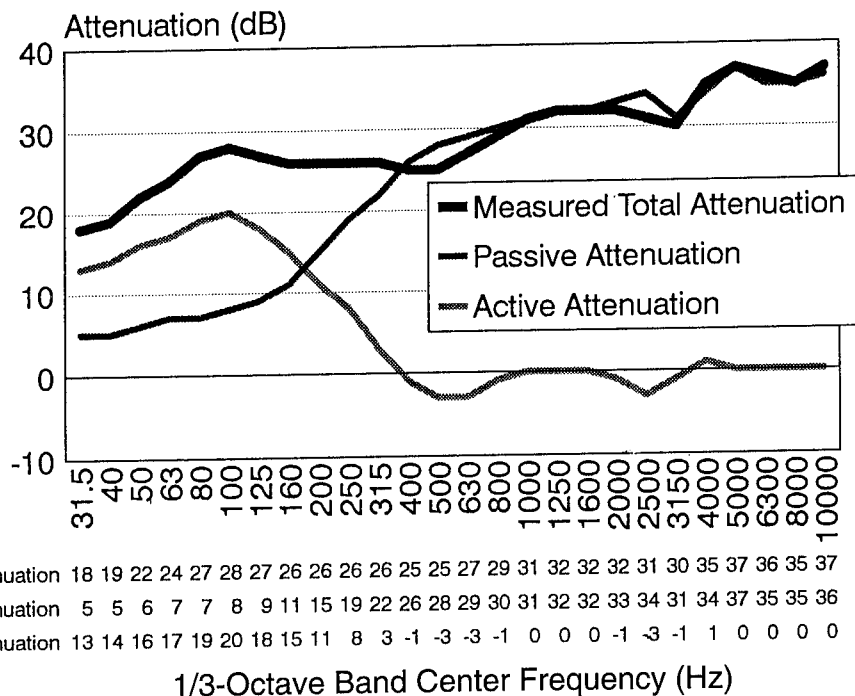


Figure 28-5 Active, passive, and total attenuation for the Peltor ANR 7004 headset.

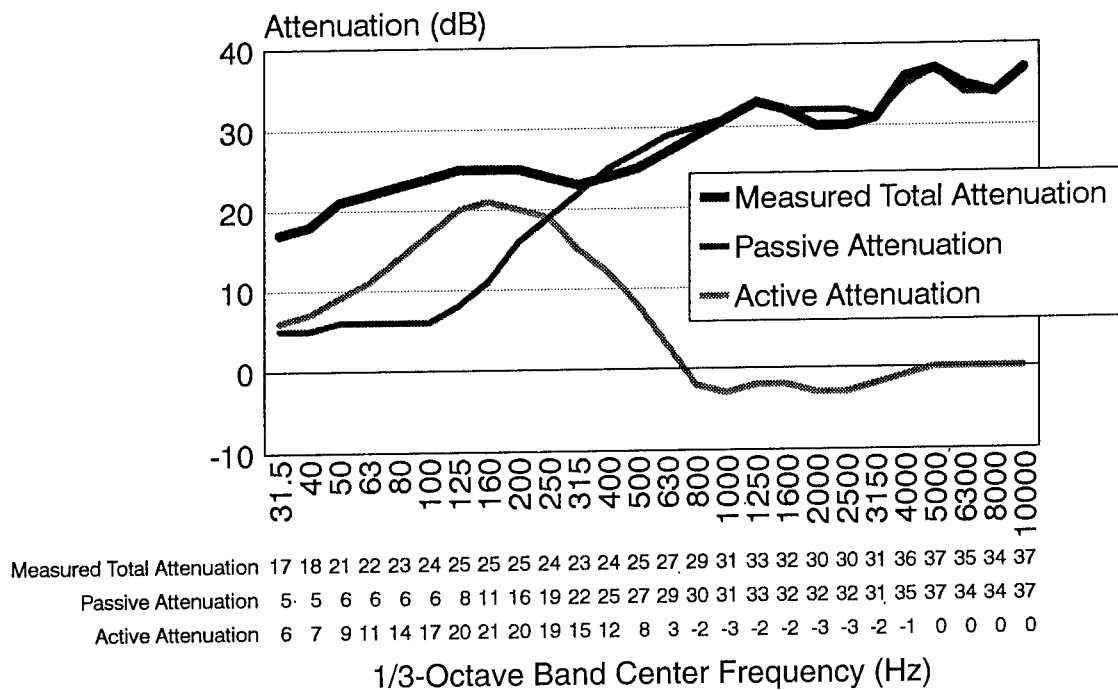


Figure 28-6 Active, passive, and total attenuation for the Sennheiser HMC 200 ANR headset.

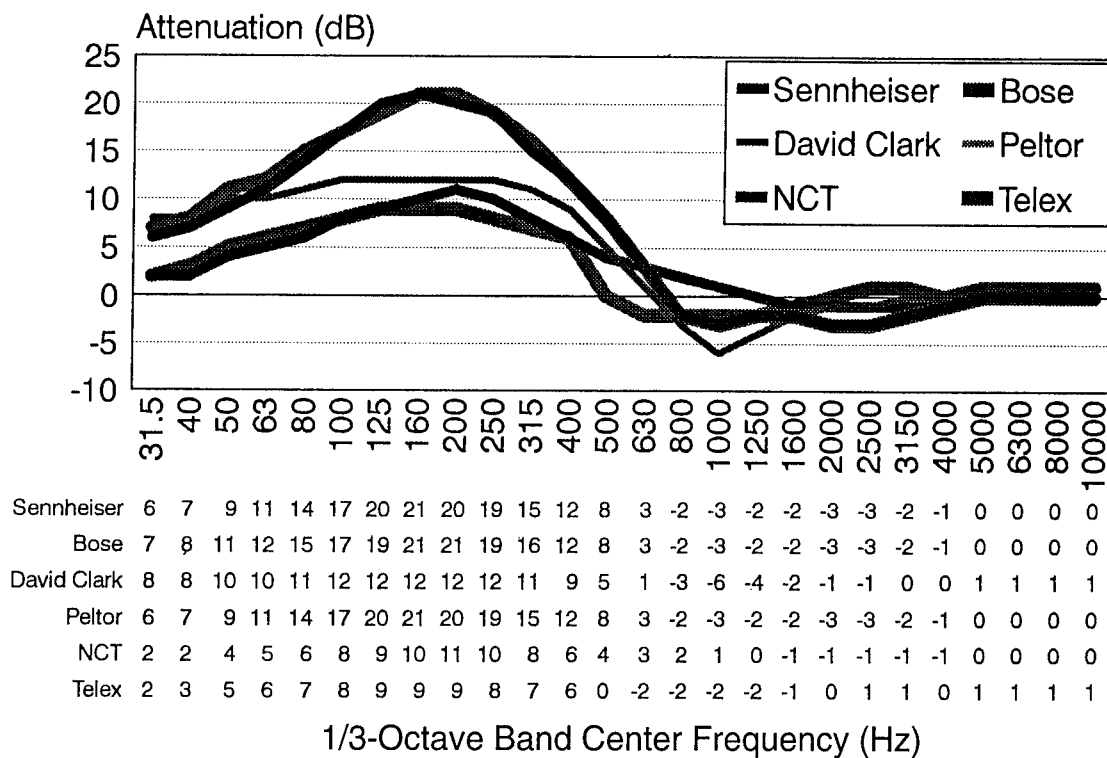


Figure 28-7 Active attenuation for six ANR headsets.

active attenuation begins at 13 dB at 31 Hz and peaks at 20 dB at 100 Hz, falling to 0 dB between 315 and 400 Hz. Figure 28-6 shows the performance of the Sennheiser HMC 200 ANR headset. The passive attenuation is 5–6 dB up to 100 Hz. The active attenuation begins at 6 dB at 31 Hz, peaking at 21 dB at 160 Hz, and declining to 0 dB between 630 and 800 Hz.

The active noise attenuation at 105 dB noise of the six ANR headsets is summarized in Figure 28-7. There is significant range of active attenuation and bandwidth. Figures 28-8–28-11 show the active attenuation for the Sennheiser, Bose, Peltor, and David Clark headsets, respectively, in 105 versus 115 dB levels of noise. The Sennheiser and Peltor ANR headsets demonstrate a decrease of approximately 3 dB in maximum active attenuation and a frequency shift of approximately two-thirds of an octave due to the 115 dB noise level. The Bose ANR active attenuation shows basically no effect due to the 115 noise level.

The David Clark ANR (Figure 28-11) shows about a 2 dB decrease in active attenuation at the 115 dB noise level.

Figure 28-12 summarizes the total attenuation of the six ANR headsets at the 105 dB noise level. The passive attenuation at both 105 and 115 dB of the three passive headsets is shown in Figures 28-13–28-15. The passive attenuation was equal within normal experimental variance for all three passive headsets at both the 105 and 115 dB noise levels. The Sigtronics S-20 headset (Figure 28-13) attenuation was 8 dB at 31 Hz to 36 dB at 8 kHz. There was a dip in the passive attenuation of approximately 6 dB from 1 to 2 kHz. The Telex 2000E passive headset (Figure 28-14) attenuation was 2–3 dB from 31 to 63 Hz, increasing to a maximum 36 dB at 5 kHz. The David Clark H10–60 passive headset (Figure 28-15) was 3 dB at 31 Hz to a maximum 34 dB at 5 kHz. Two of the better ANR headsets' total attenuation is summarized with the three passive headsets' attenuation in Figure 28-16.

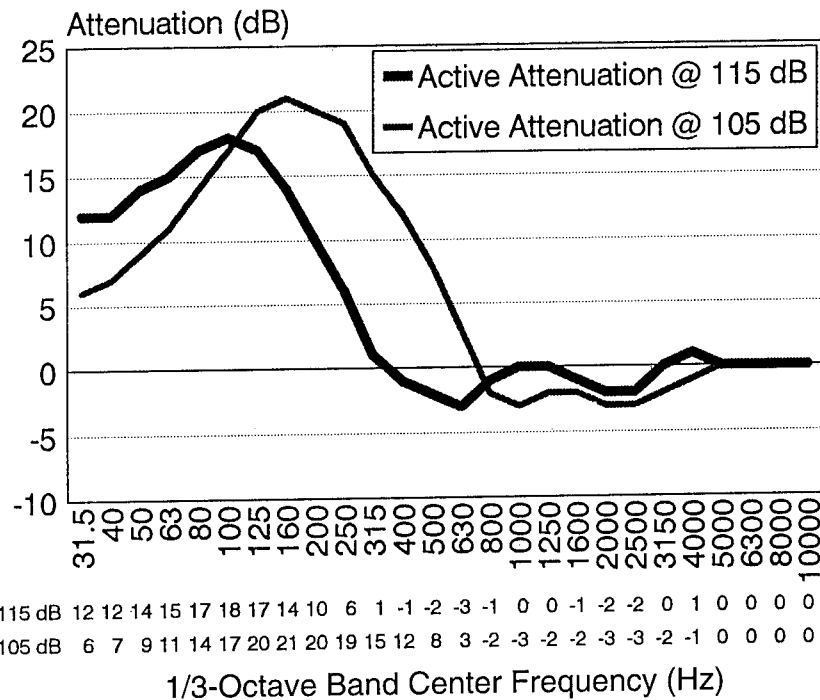


Figure 28-8 Active attenuation at 105 dB and 115 dB sound pressure levels for the Sennheiser HMC 200 ANR headset.

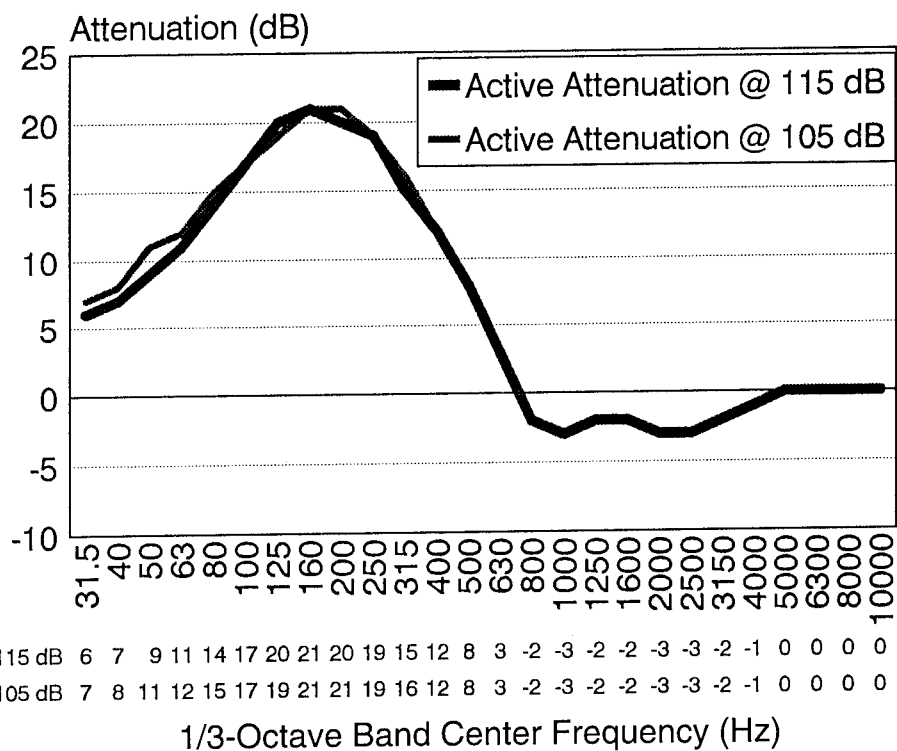


Figure 28-9 Active attenuation at 105 dB and 115 dB sound pressure levels for the Bose Aviation headset.

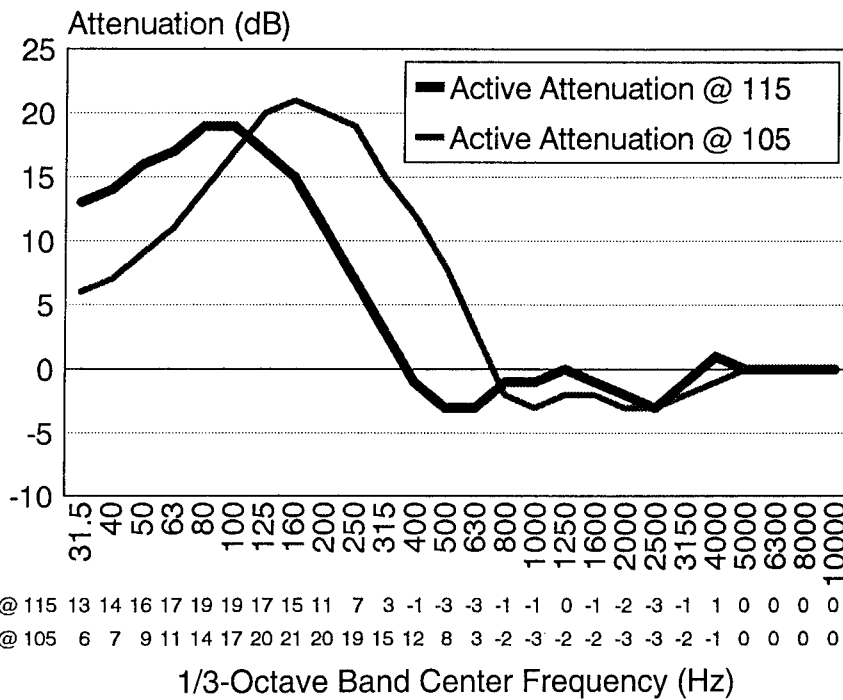


Figure 28-10 Active attenuation at 105 dB and 115 dB sound pressure levels for the Peltor ANR 7004 headset.

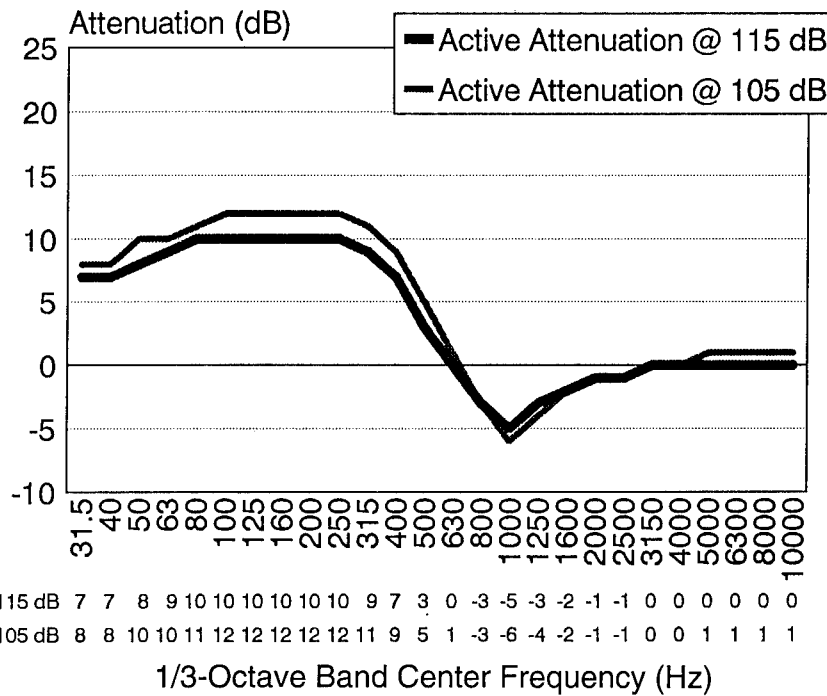


Figure 28-11 Active attenuation at 105 dB and 115 dB sound pressure levels for the David Clark ANR headset.

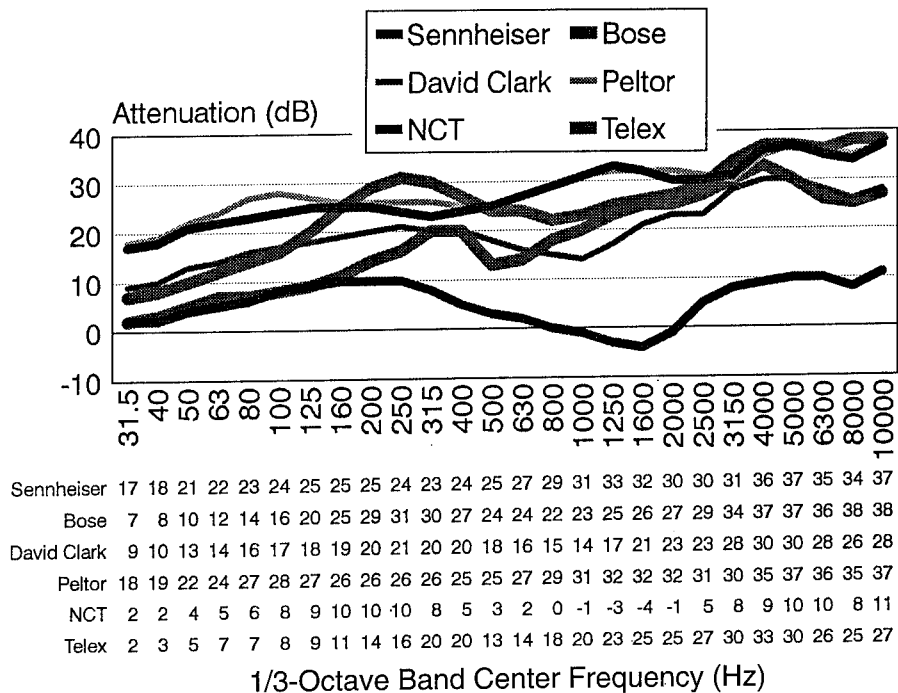


Figure 28-12 Total attenuation for six ANR headsets.

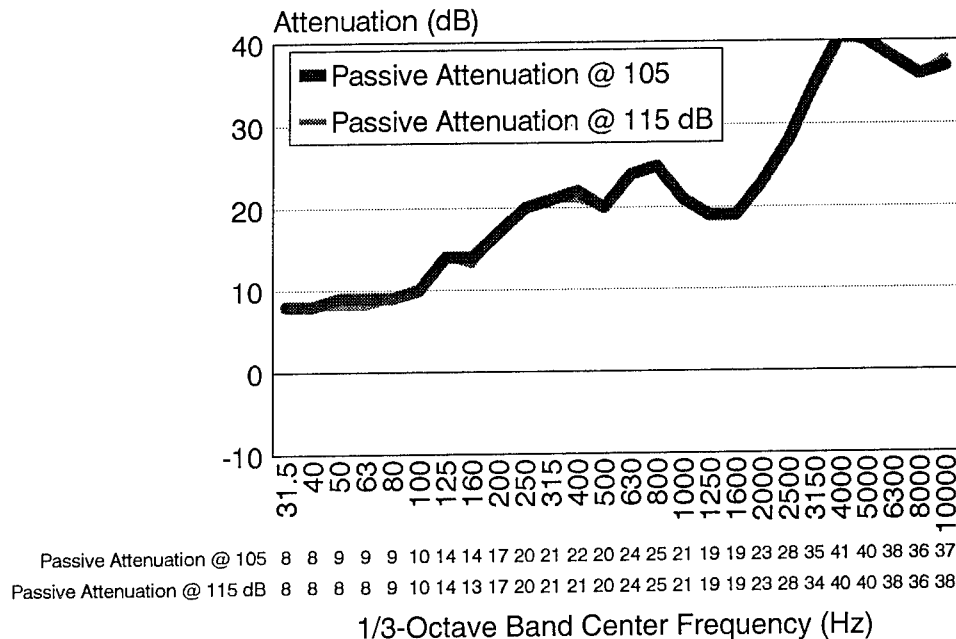


Figure 28-13 Passive attenuation at 105 dB and 115 dB for the Sigtronics S-20 passive headset.

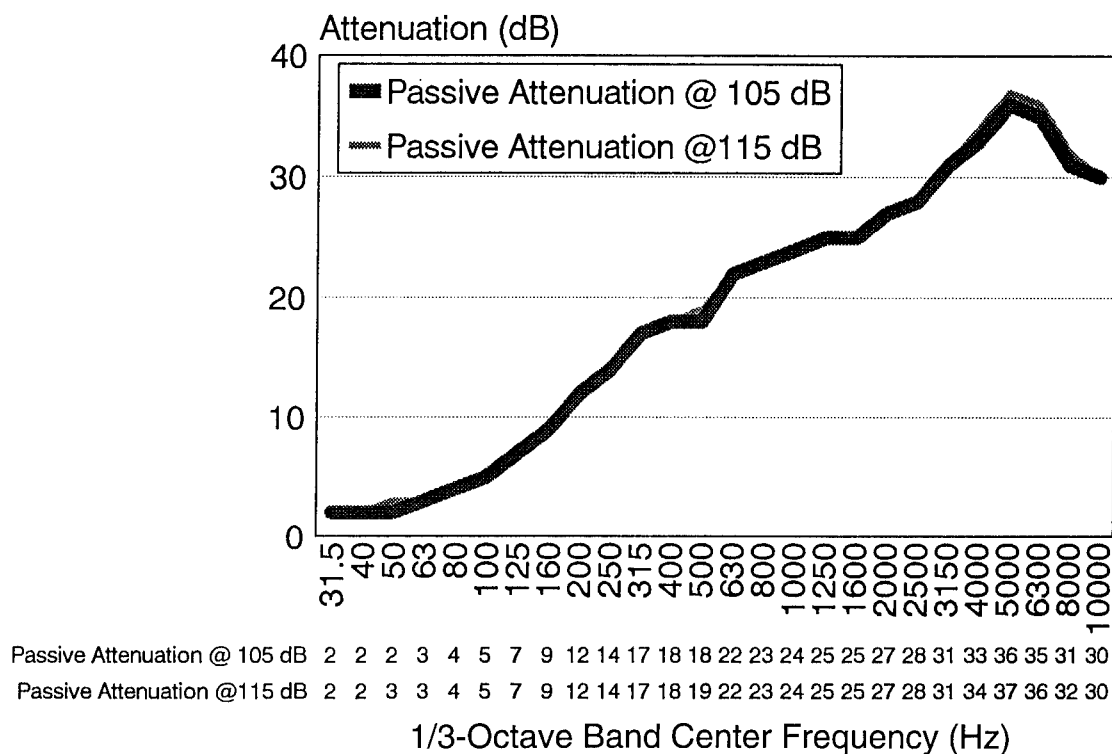


Figure 28-14 Passive attenuation at 105 dB and 115 dB for the Telex passive headset.

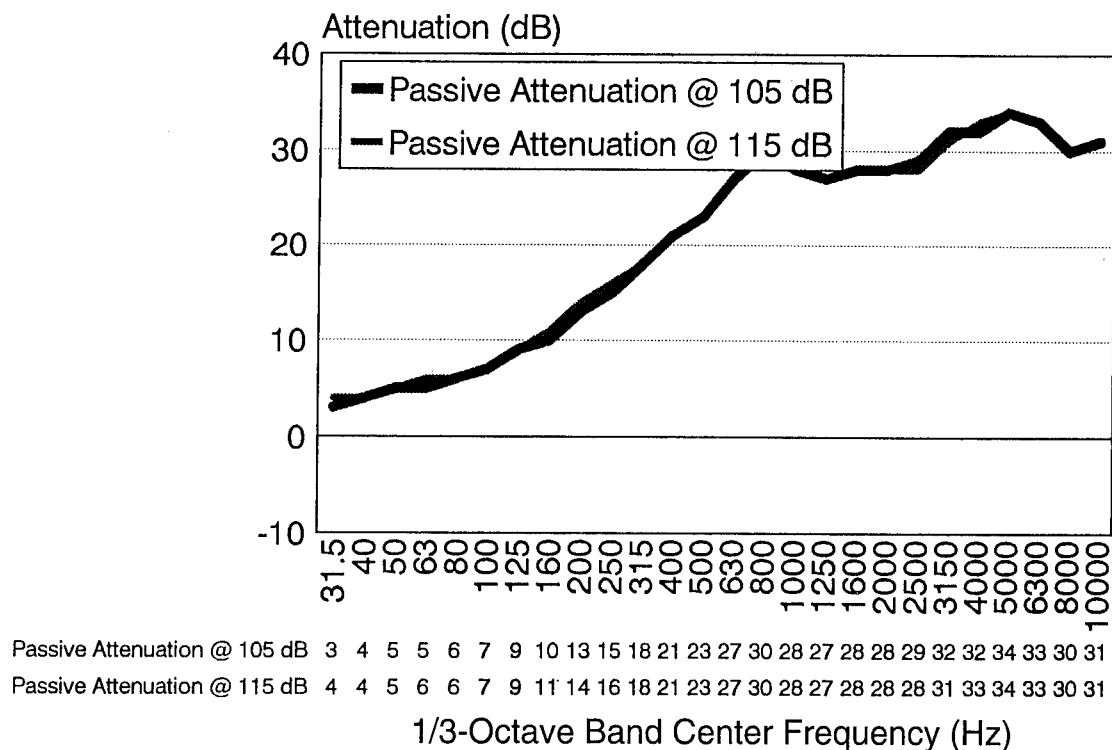


Figure 28-15 Passive attenuation at 105 dB and 115 dB for the David Clark passive headset.

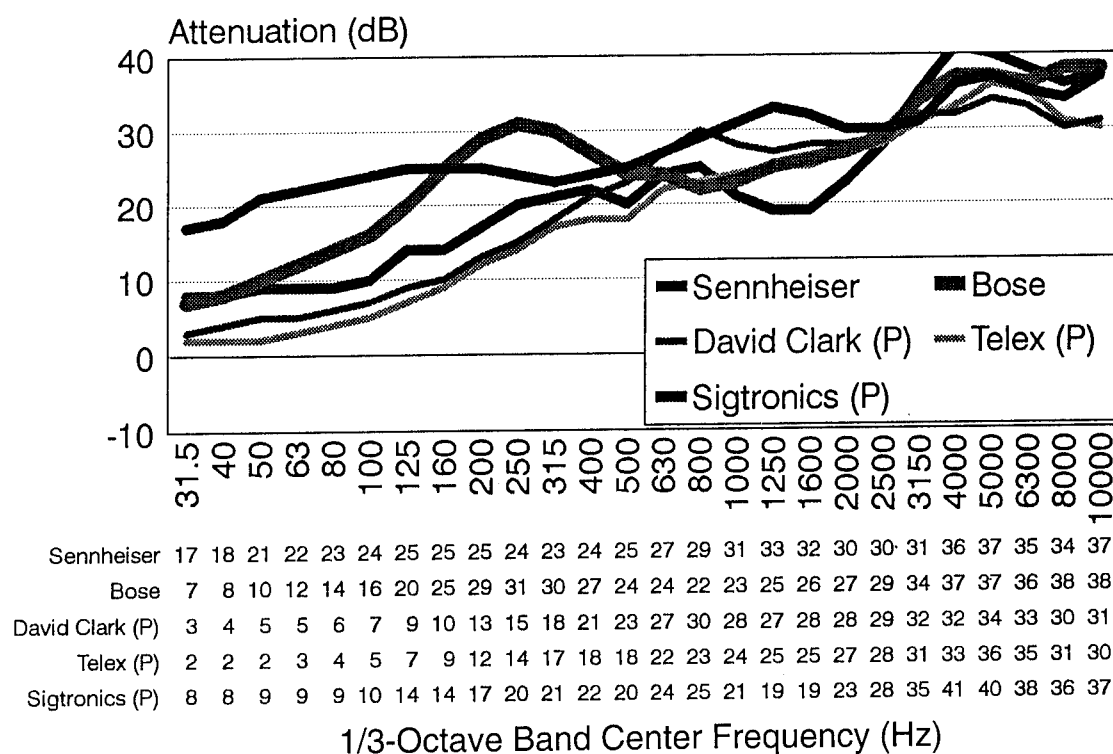


Figure 28-16 Comparison of total attenuation for two ANR headsets versus three passive headsets.

Using the procedures described in the Method section, estimated noise-induced hearing loss was calculated for a standard passive headset installed in a helmet and for one of the better ANR headsets installed in a helmet. The noise environment used was a broadband noise found in jet training aircraft. Figures 28-17 and 28-18 show the estimated noise-induced hearing loss for two populations, instructor pilots and ground crew, respectively, for the 10th through the 90th percentiles.

Discussion

The total attenuation of the ANR headsets ranged from much poorer than the passive headsets to better than the passive headsets. The NCT NB-DX headset is an example of a headset that could be used in a nonhazardous noise environment (less than 85 dBA) for qualitative improvements for listening, but should not be used to provide basic hearing protection. The David Clark ANR headset provided

adequate passive and active attenuation but was not one of the better performing ANR headsets. The Telex ANR 4000 headset active attenuation was 0 dB by 500 Hz. This fact along with the typical sound reinforcement in the higher frequencies generated a dip in the total attenuation from 500 Hz to 1 kHz. This range of performance is important to potential users of ANR headsets. The Bose Aviation headset demonstrated low passive attenuation at the low frequencies and good passive attenuation at the high frequencies. The active attenuation was one of the best, and tolerated increased noise levels were better than any of the other ANR systems in this study. The Peltor ANR 7004 headset and the Sennheiser HMC 200 ANR headset are discussed together because physical examination revealed them to be nearly identical except for the name engraving. The Peltor/Sennheiser ANR headset has the best low-frequency passive attenuation of any of the ANR headsets and was one of the best in active attenuation. The total at-

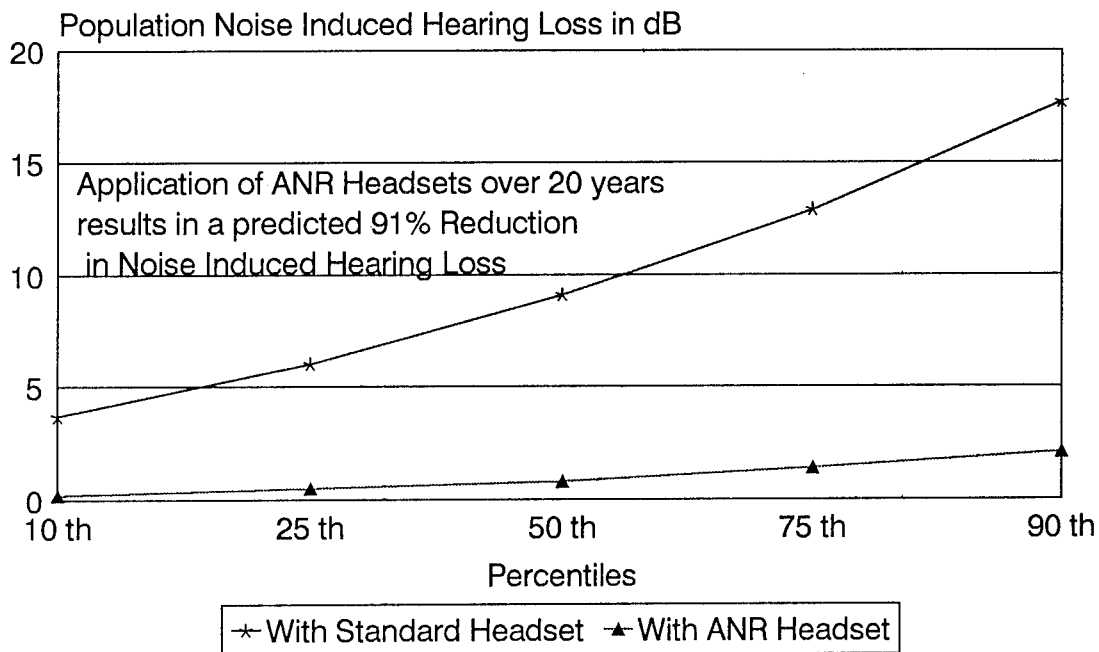


Figure 28-17 Estimated noise-induced hearing loss for an ANR and a passive headset used by a groundcrew person.

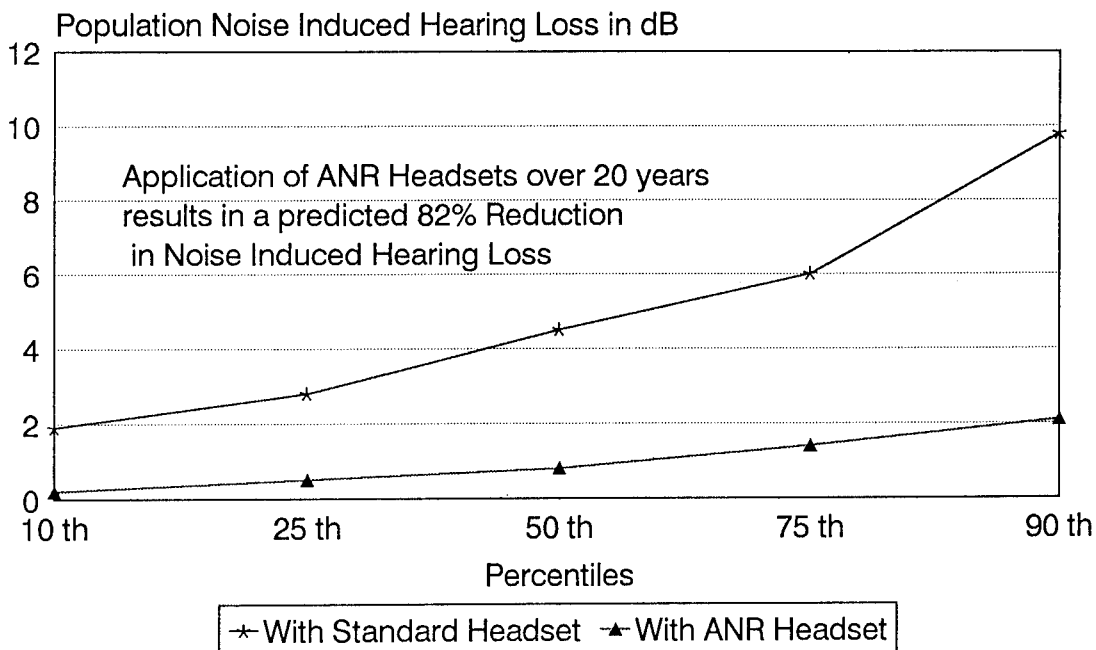


Figure 28-18 Estimated noise-induced hearing loss for an ANR and a passive headset used by an instructor pilot person.

tenuation for the Peltor/Sennheiser was the best in this study. However, the response of the Peltor/Sennheiser ANR headset to increasing noise levels was different than any of the other headsets. A decrease in active performance is expected at some noise level with any ANR headset. The Peltor/Sennheiser ANR headset, however, showed not only a decrease, but also a frequency shift in active attenuation. The passive headsets showed a range of performance that can be expected from a cross section of manufacturers.

Currently, there is no American national standard or international standard for the measurement of the noise attenuation of ANR headsets. Furthermore, ANR manufacturers typically do not provide active attenuation data for their ANR products. The better performing ANR headsets do provide a benefit over the traditional passive attenuating headsets in this study. When total attenuation is considered and the A-weighted noise exposure at the ear (under the protector) is being controlled by noise at 500 Hz and below, ANR headsets should be considered in a cost/benefit analysis. Little advantage can be gained by using ANR headsets in noise environments with the predominant energy above 1000 Hz.

ANR headsets are most beneficial in environments dominated by low-frequency noise and in some cases broadband noise. The reductions in the A-weighted noise level at the ear can be 10–20 dB in many cases. In these situations the predicted effects of ANR headsets on noise-induced hearing loss are dramatic. For the broadband case the reductions in estimated noise-induced hearing loss due to the application of ANR headsets were 82 and 91% for the instructor pilot and ground-crew populations, respectively.

Summary

Most ANR headsets have targeted specific applications involving high-intensity, low-frequency noise. This study provides potential users of ANR headsets with current performance data of ANR headsets and with performance of a sample of passive headsets. The data demonstrate that some ANR headsets are effective in providing improved low-frequency attenuation. The end benefit is reduced overall noise exposure and decreased risk of noise-induced hearing loss. However, there are applications where ANR headsets do not provide any significant benefit. Conventional passive earplugs and earmuffs, singly and in combination, can provide adequate attenuation for most low- and moderate-level noise environments. However, there are many industrial noise environments that can profit from the use of ANR headset technology. The hearing conservationist may wish to consider the information provided herein to determine if the capabilities of ANR match the characteristics of noise to which employees are exposed and if it provides an advantage in noise attenuation and decreased incidence of noise-induced hearing loss.

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Chapter 29

International Review of Field Studies of Hearing Protector Attenuation

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When a manufacturer designs a hearing protection device (HPD), a hearing conservationist specifies its use, or a purchaser selects it for a particular application, one question foremost in their minds is just how much noise reduction (also called attenuation) the device will provide. Until the middle 1970s this question was always answered using test data obtained under closely controlled conditions in a laboratory setting. The degree to which such data corresponded with actual use, often called "real-world" performance, was not only unanswered, but also rarely if ever asked. This changed in the latter part of the 1970s as studies began to appear in the literature that presented the results of attenuation experiments conducted in the real world. Subjects in the studies were persons actually wearing HPDs for protection from occupational noise.

Although there have been at least 22 reported studies worldwide since 1975, that have examined real-world attenuation of HPDs,¹⁻²² and a review paper published in 1983 that summarized the data from the 10 studies available at that time,²³ controversy still exists concerning real-world attenuation. The debate centers around the extent of the divergence between values measured in the laboratory under ideal and commonly standardized conditions and those values observed in the real world, and how to best use laboratory data to predict real-world performance for particular applications. Herein we update Berger's 1983 summary, and provide a definitive picture of the real-world attenua-

tion of hearing protectors circa 1994. We also present representative laboratory test data so that its validity (or realism), that is, the accuracy with which it predicts real-world performance, can be assessed.

Estimation of effective protected noise exposures when hearing protectors are worn not only requires valid HPD attenuation data, but also accurate noise exposure measurements as well as a suitable computational scheme with which to utilize such values. Noise measurements and predictive methods are not the subject of this chapter, but the results of such computations are of course heavily influenced by the attenuation data described herein. A recently issued ISO standard²⁴ describes three computational approaches. The reader is also encouraged to review Lundin²⁵ and Waugh²⁶ for background analyses and discussion.

Real-World Data Sample

The first reported data on field performance of HPDs appeared in 1975.¹⁹ Since then, we are aware of at least 21 additional studies available worldwide.^{1-18,20-22} The total data base comprises results from over 90 different industries, in seven countries (Argentina, Canada, Finland, Germany, Netherlands, United Kingdom, and United States) with a total of approximately 2900 subjects.

Field measurements have been conducted by independent researchers, government-sponsored investigators, and staff employed by the industries supplying the data. In all cases, the test subjects were workers or mili-

tary personnel exposed to noise who were tested in most cases while wearing their own HPDs.

The facilities that have been studied most likely represent the better hearing conservation programs in existence. This presumption is based upon the increased likelihood of finding higher quality programs among companies and organizations interested in and choosing to participate in the complicated, time-consuming, and costly research of the type required for real-world evaluations. In fact in at least two of the more recent studies, the locations were selected specifically because the authors believed them to be exemplary.^{9,18}

Candid Versus Scheduled

Subject participation in field studies has been based upon either *candid* selection or *scheduled* testing. Candid studies are the type in which subjects know that their work site is under investigation and that they will be asked to participate, but they do not know when. The researcher selects them without warning and then escorts them to the test facility while monitoring them to assure that they do not readjust the fit of their HPDs. Scheduled tests describe situations in which either the subjects are notified in advance and asked to come to the test facility bringing their HPDs with them to fit at the time of the test, or may be of the type where subjects are fitted with earmuffs instrumented with small microphones to measure the interior and exterior noise levels while they wear their HPDs during the work day.

At face value it might seem that candid studies would provide a truer picture of actual real-world usage than would scheduled studies. For the scheduled test it would appear axiomatic that the subject would purposely fit the device differently, a better fit because the testing is under the watchful eye of the experimenter or the subject wants to look good; a poorer fit because the subject wants to sabotage the test results.

For four of the insert HPDs evaluated, there were enough studies of both types to examine

the effect of scheduling. Although for three of the earplugs, the scheduled tests tended to show higher attenuation values by a few decibels in terms of the Noise Reduction Rating (see Real-World Data and Metrics Utilized in This Report), the candid and scheduled data agreed within a few tenths of a decibel for the device on which the largest number of studies were conducted (E·A·R®/Decidamp earplugs, see Table 29-1). The foam earplug is also the one for which attenuation can be varied most easily and dramatically by subject-insertion method, and thus would have been anticipated to be the one most susceptible to bias on the part of the test subjects. For the remainder of this chapter, the data from both the candid and scheduled procedures will be pooled for analysis and discussion.

REAT Procedure

Two principal methods have been used to measure real-world attenuation: real-ear attenuation at threshold (REAT) and microphone in real ear (MIRE). For a complete discussion see Berger.²⁷

REAT can be conducted with all types of hearing protectors as long as the facility presents the test signals in a sound field, even if the sound field is only that found in a small portable audiometric booth. However, because of potential background-noise masking problems, as well as cost and convenience considerations, it is generally easiest to conduct field REAT measurements using large circumaural earcups with built-in loudspeakers to generate the requisite sound field for the open and occluded measurements. Even so, masking of low-frequency open thresholds can occur. This will lead to underestimates of REAT. With headphone-based REAT procedures only earplug type HPDs can be evaluated.

Typically, under field application of REAT, a subject is first tested with the HPD in place as it was worn on the job, followed by an open threshold. The difference is the presumed real-ear attenuation. Because of possible learning effects between the occluded and open audiograms, the open threshold values

may be spuriously improved by a few decibels simply due to better test-taking skills on the second test, and hence the REAT increased. This potential error, which can lead to overestimates of attenuation, is in the opposite direction to that caused by background-noise masking effects noted above.

An interesting alternative REAT procedure, the reference-earmuff method, was utilized in one study to measure earmuff and semiaural device attenuation.²¹ The authors selected it because they were concerned about room noise producing masking of the open ear thresholds, which can easily occur under field test conditions. They sought a method like that of headphone-based REAT in which thresholds are always measured inside noise-excluding earcups. But, they wanted to be able to test earmuffs, an option that would be precluded by a headphone-based procedure.

The solution was to establish both real-ear attenuation and the occluded threshold levels for test subjects wearing a reference earmuff in the laboratory. In the field, measurements were taken of the occluded thresholds (no unoccluded values were measured in the field) for both the product being field tested (candid subject fit) and the reference earmuff (experimenter-supervised fit). The attributed attenuation was then calculated as the laboratory attenuation of the reference earmuff plus (or minus) the difference between the occluded thresholds of the reference earmuff and the test HPD, under field conditions. The accuracy of this method is strongly dependent upon the particular attenuation values selected for the reference earmuff, and the presumption that the attenuation of the reference earmuff achieved by the field test subjects closely approximates the values found in the laboratory using a different panel of listeners.

MIRE Procedure

The MIRE procedure, as implemented in field studies, consists of mounting small microphones inside and outside a hearing protector while it is worn by an employee on the job. The "test noise" is the actual noise to which

the employee is occupationally exposed. The attenuation values that are reported can either be the differences in spectral sound pressure levels recorded by the two microphones, or the differences in time-averaged values of the A-weighted sound pressure levels (i.e., noise doses).

Because of the intrusiveness of mounting interior and exterior microphones, field MIRE measurements, unlike REAT, can only be applied to circumaural HPDs. The advantage of MIRE is that it allows a continuous monitoring of the noise levels, and an objective measurement independent of the subjects' ability to take an audiogram. The disadvantage is the limitation of being able to only test earmuffs, and the fact that the experimenter and the procedure may directly influence the subjects' use of the HPDs. This may enhance attenuation as a result of the additional attention the wearer receives, or reduce attenuation if the cabling and microphones interfere with the earmuff's ability to properly seal and block noise.

MIRE is best measured via an insertion loss (IL) protocol in which the sound levels in the canal are measured with and without the HPD in place. This directly corresponds to the paradigm inherent in REAT, and is how MIRE is normally implemented in the laboratory. However, for practical reasons the implementation of MIRE in field studies is always done with interior (canal-, or concha-measured) and exterior noise levels simultaneously recorded to yield a noise reduction (NR) value instead of an IL value.

In the NR protocol the reference microphone is the exterior microphone. It records lower sound levels than the ear canal mounted reference microphone used in the IL method, because it does not benefit from the amplification of the transfer function of the open ear. Thus, the difference between the occluded measurement (interior microphone) and the open measurement (exterior microphone) is less than occurs with IL procedures. Because most authors do not correct their field-measured MIRE values, they tend to provide low attenuation estimates, by about 5 dB or so, at and above 3 kHz.

Laboratory Data Base

For purposes of comparison to the field data summarized herein, various graphs and tables also provide the associated labeled test data based upon manufacturers' published North American laboratory results.

Laboratory testing of HPDs in North America is conducted in conformance with standards promulgated by the American National Standards Institute.^{28,29} The procedures call for determining "*optimum performance values which may not usually be obtained under field conditions*" (author emphasis). Optimum performance values, as opposed to estimated real-world values, have historically been specified for laboratory testing because US standards groups have felt that those values could be more consistently replicated, and were useful for rank-ordering HPDs. However, current data as described herein, and reported by Berger³⁰ suggest otherwise. Nevertheless, ANSI S3.19/S12.6 type data *are the only standardized values* that regulators and manufacturers in the United States currently have available for labeling and informational purposes.

In Europe, testing has been conducted according to ISO 4869.³¹ The procedure is essentially the same as in the American standards, but the subject fitting practices are described somewhat differently and have typically been interpreted in ways that yield lower laboratory attenuation values, especially for insert-type HPDs, than do the tests reported by manufacturers on the other side of the Atlantic ocean.³² Sample European data appear in selected octave-band charts to follow.*

Real-World Data and Metrics Utilized in This Report

The data reported in the 22 field studies are mean attenuation and standard deviation values. It is those data that are graphically

presented in the accompanying figures. The authors' values have been utilized as reported. If they measured NR and failed to correct the values to estimate IL, then the NR measures were reported. Only in one instance were the raw data adjusted.⁹ In that case background noise measurements were available to confirm that the low-frequency open thresholds were masked, spuriously reducing the measured real-ear attenuation. The values were mathematically corrected.³³ In some cases where authors reported data at fewer frequencies than required for computation of the Noise Reduction Rating (NRR), the NRR was estimated based upon empirical relations between attenuation at key octave bands and overall attenuation.³⁰

The NRR was selected as a simplified single-number metric of an HPD's overall real-world attenuation, because it is standardized for labeling purposes,³⁴ it has been in use for over a decade, and it is well known in the hearing conservation community. For a given set of data and a given theoretical percentage of the population protected, the NRR is approximately 3 dB less than the Single Number Rating (SNR), the single-number metric defined in the recently released international standard, ISO 4869-2.²⁴

The *labeled* NRRs were computed per the US Environmental Protection Agency, by subtracting a 2-standard deviation (SD) correction from the mean attenuation values in order to estimate the minimum noise reduction theoretically achieved by 98% of the laboratory subjects (NRR_{98}). The field data were computed in the same manner except that only a 1-SD correction was included, thus estimating the minimum attenuation achieved by 84% of the actual wearers (NRR_{84}).

The 2-SD deduction required in the labeled NRRs (i.e., NRR_{98S}) causes many field-measured NRRs to become negative numbers. A smaller 1-SD subtractive correction can avoid this problem. A 1-SD correction is also more in keeping with the practices of most of the non-North American community. With more realistic test data (i.e., larger SDs) it provides a better balance between adequately

*In this report, European data consist of results taken from manufacturers' European published data sheets, as well as data from the Karolinska Institute, Stockholm, Sweden.

protecting a majority of wearers and avoiding overprotection of a minority. Additional justification for use of a 1-SD correction stems from consideration of the heightened impact of outliers when 2-SD corrections are used, the reduction of between-study variability when only 1 SD is accounted for, and the variability of the susceptibility of individuals within a population to noise-induced hearing loss.³⁵

The issue of whether field attenuation data are suitably normal to apply Gaussian-based SD corrections was examined by comparing estimates of the actual 84th percentile, to those obtained by subtracting 1 SD from the mean attenuation values. The data consisted of five 50-subject, and one 100-subject, subject-fit attenuation data sets, for four earplugs and two earmuffs. Both over- and underestimates of the true 84th percentile occurred, with the average error being 0.5 dB and the maximum error 3.1 dB. Examination of the same question using the real-world data of previous reports,^{8,10} leads to errors of typically <2 dB, with the maximum difference between the 84th percentile and a 1-SD estimate of that value, being 4.2 dB.

Tabular Overview

The authors were able to gather from the 22 reports nearly 100 sets of data on approximately 40 different devices, each data set being defined as the attenuation at one or more frequencies for one HPD for one group of subjects. The results for all of the devices, sorted into five insert and two circumaural categories (excluding three HPDs which did not easily fit into any of the groupings), and averaged across studies, are summarized in Table 29-1. Individual devices and/or subcategories were selected so that similar products were assembled together, and so that the number of subjects for each subcategory was greater than 30. Another requirement for a device to be individually listed in a row was that published US laboratory test data had to be available for inclusion in the data set. Data from 2879 subjects out of a total possible population of 2945 subjects are included in Table 29-1.

For each row, the number of studies contributing data as well as the total number of subjects are shown, along with the real-world NRR_{84} averaged across the group of studies noted for that row. The labeled NRR_{98} based upon manufacturers' North American published laboratory test results is also reported. The last column provides the relationship between the real-world NRR_{84} and the labeled NRR_{98} as a percentage. The field NRRs for earplugs yield only 5–52% of the labeled values (averaging about 25%), and for earmuffs, from 47 to 76% (averaging about 60%).

Representative Octave-Band Results

Representative field-performance data are presented in Figures 29-1–29-8, to illustrate the types of octave-band results observed in the various studies. The data include the results for: the earplug shown to provide the least attenuation under real-world conditions; an earplug with average real-world attenuation and very low interstudy variability; the earplug with the highest average real-world attenuation; and the earmuff on which the most real-world studies have been conducted. Figures 29-1, 29-3, 29-5, and 29-7 provide the individual data from each of the studies, and 29-2, 29-4, 29-6, and 29-8 present the data averaged across real-world studies with a comparison to both North American published manufacturers' data and representative European test data.

The results indicate that depending upon interpretation of the relevant test standard and implementation of subject selection, training, and fitting practices by the researcher, laboratory data may provide a more valid (European) or less valid (American) estimation of field performance. An American accredited standards working group, S12/WG11 (Field Effectiveness and Physical Characteristics of Hearing Protectors) as well as the National Institute for Occupational Safety and Health are cognizant of the problem and are currently conducting research and developing a new laboratory test method to address these issues.³⁶

Table 29-1 Real-World (RW) Data Summary from 22 Independent Studies

Device Type	Device	No. Studies	No. Subjects	Mean RW NRR ₈₄ (dB)	Labeled NRR (dB)	NRR ₈₄ / Labeled NRR (%)
Foam	E-A-R/Decidamp	15	633	13.4	29	46
Premolded	Ultra Fit	3	58	7.3	21	35
	V-51R	5	308	2.2	23	10
	Misc. 3-flange	2	31	4.5	26	17
	EP100	5	153	1.4	26	5
Fiberglass	Down	2	84	3.5	15	23
	POP	6	196	7.8	22	36
	Soft	3	80	4.7	26	18
	Custom	6	447	5.4	17	33
Semiaural	Sound-Ban #10 / #20	2	42	9.3	18	52
Earplugs (average)			2032	6.0	22.3	27
Earmuffs	Bilsom UF-1	2	41	16.3	25	65
	MSA Mk IV	4	89	10.8	23	47
	Peltor H9A	1	34	14.0	22	64
	Misc. Muffs	11	450	15.3	23	66
	Bilsom 2313	2	56	17.5	23	76
Cap-mounted	Hellberg No Noise	1	58	11.0	23	48
Earmuffs	Peltor H7P3E	1	36	13.0	24	54
	Misc. Cap Muffs	3	83	16.3	22	74
Earmuffs (average)			847	14.3	23.1	62
Grand average			2879	10.1	22.7	45

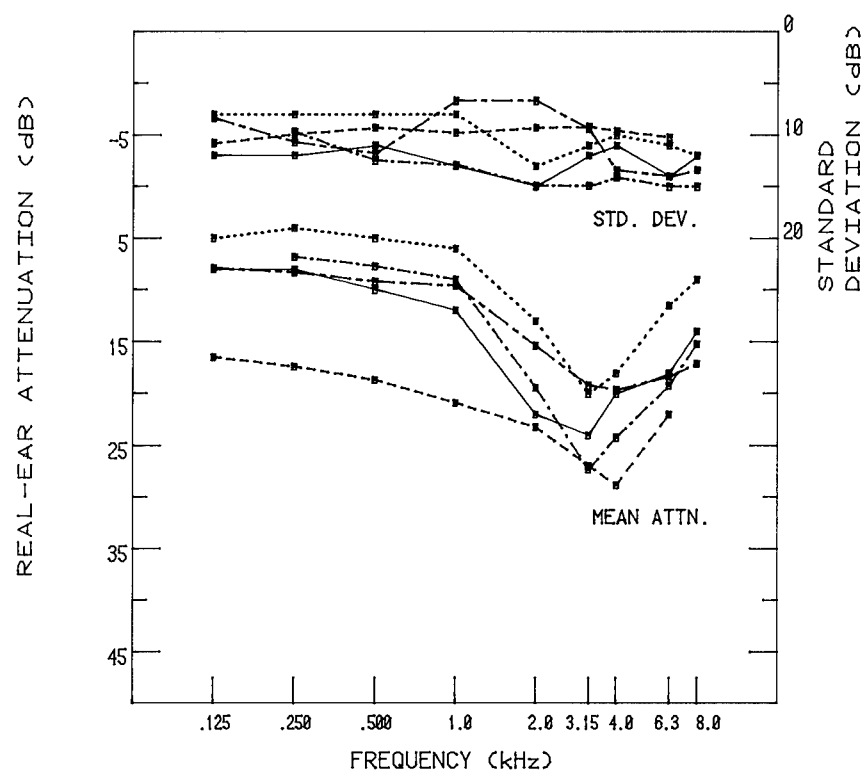


Figure 29-1 Real-world performance of the Willson EP100 premolded earplug (five studies, 153 subjects).

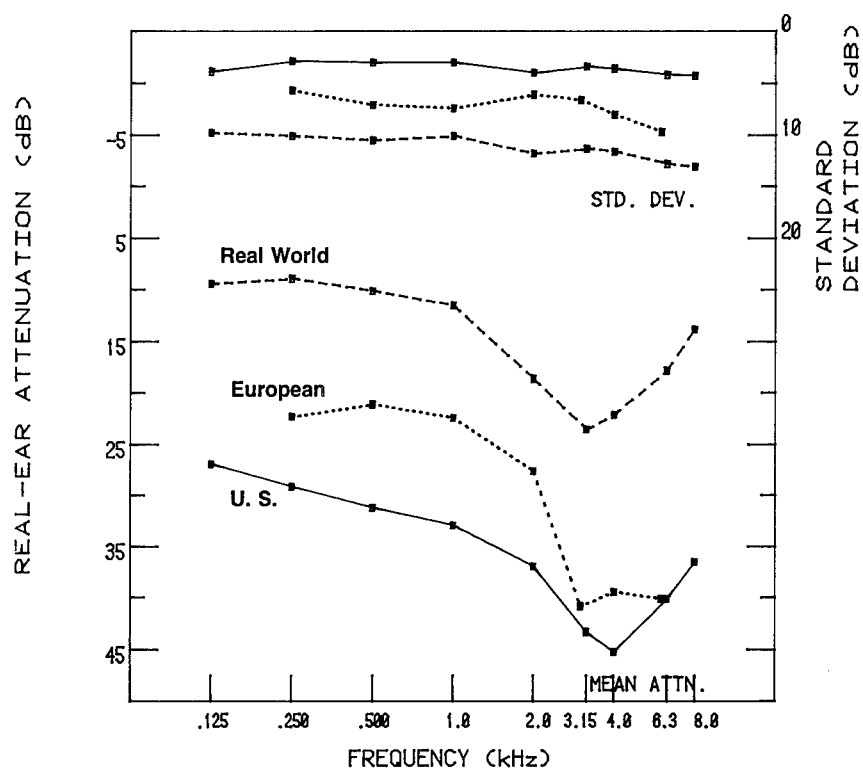


Figure 29-2 Willson EP100 earplug: real-world attenuation compared to manufacturer's US test data and European laboratory results.

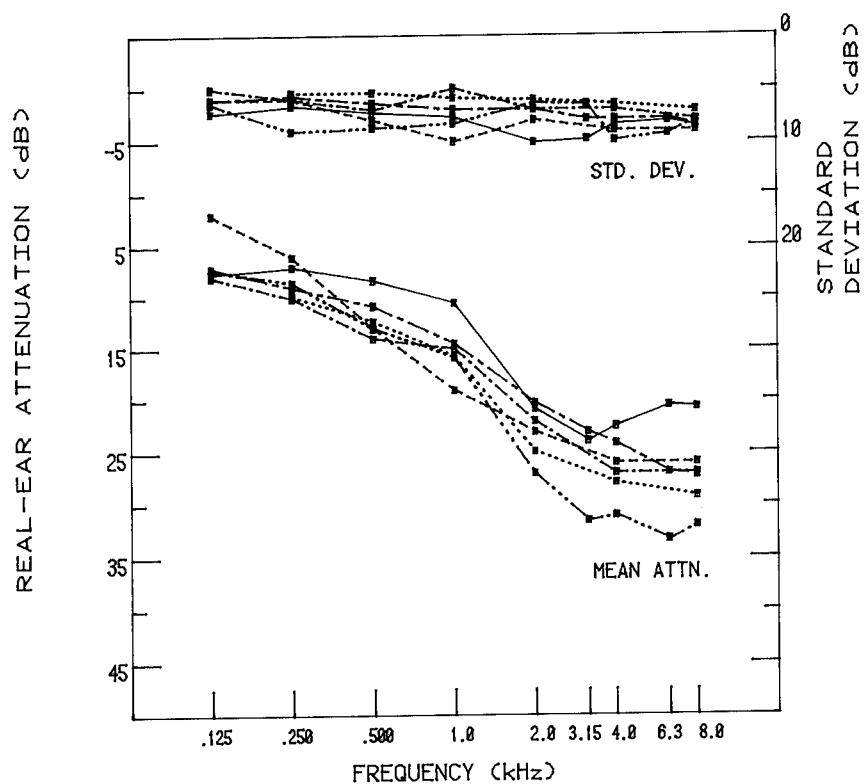


Figure 29-3 Real-world performance of the Bilsom P.O.P. sheathed fiberglass earplug (six studies, 196 subjects).

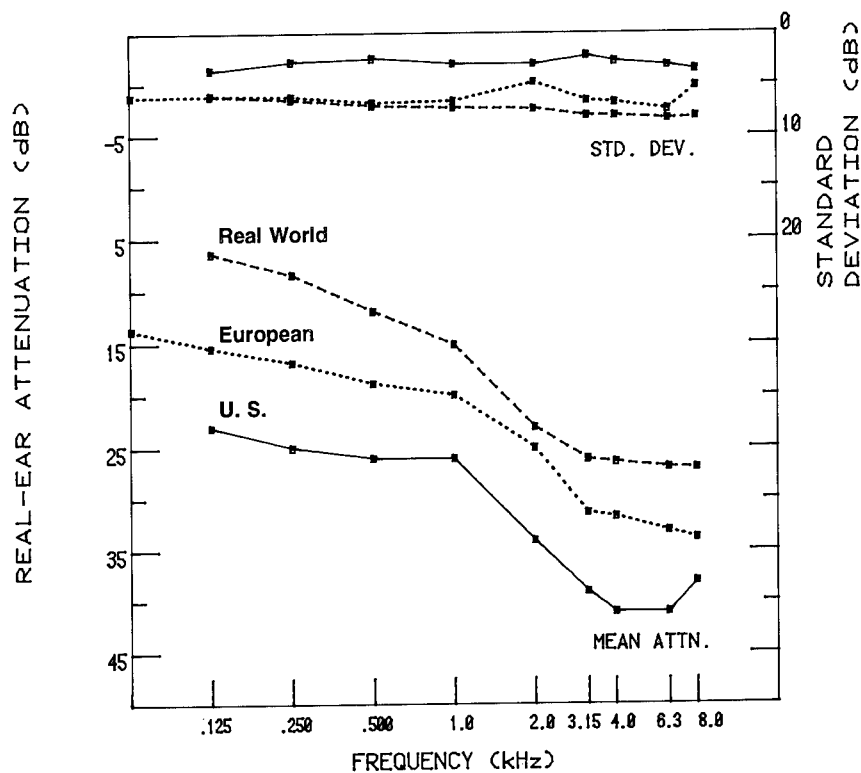


Figure 29-4 Bilsom P.O.P. earplug; average real-world attenuation compared to manufacturer's US test data and European laboratory results.

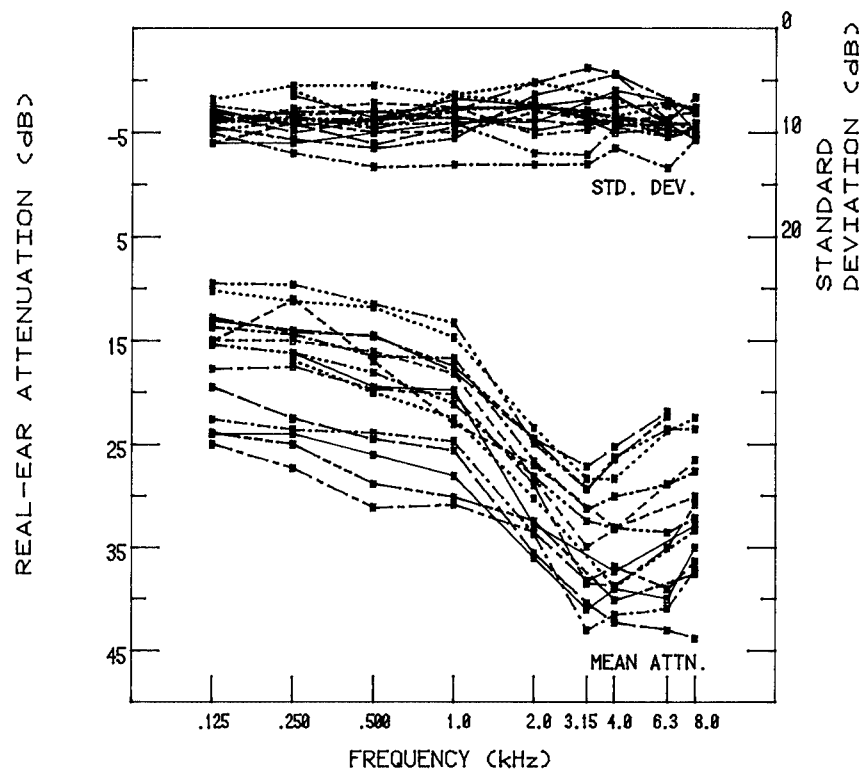


Figure 29-5 Real-world performance of the E·A·R/ Decidamp foam earplugs (12 studies, 633 subjects).

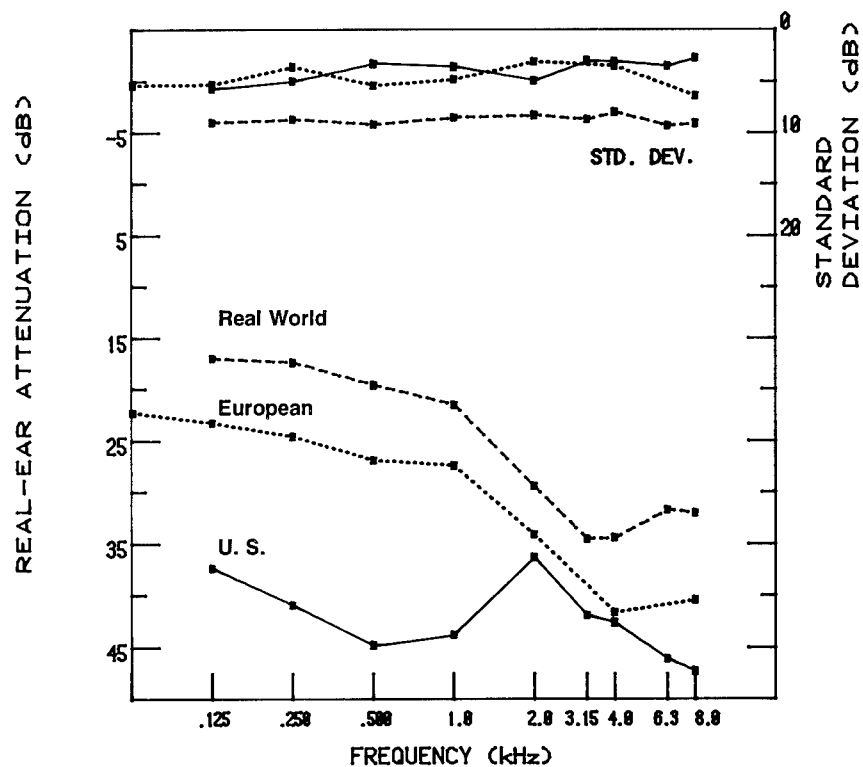


Figure 29-6 E·A·R/Decidamp earplugs: average real-world attenuation compared to manufacturer's US test data and European laboratory results.

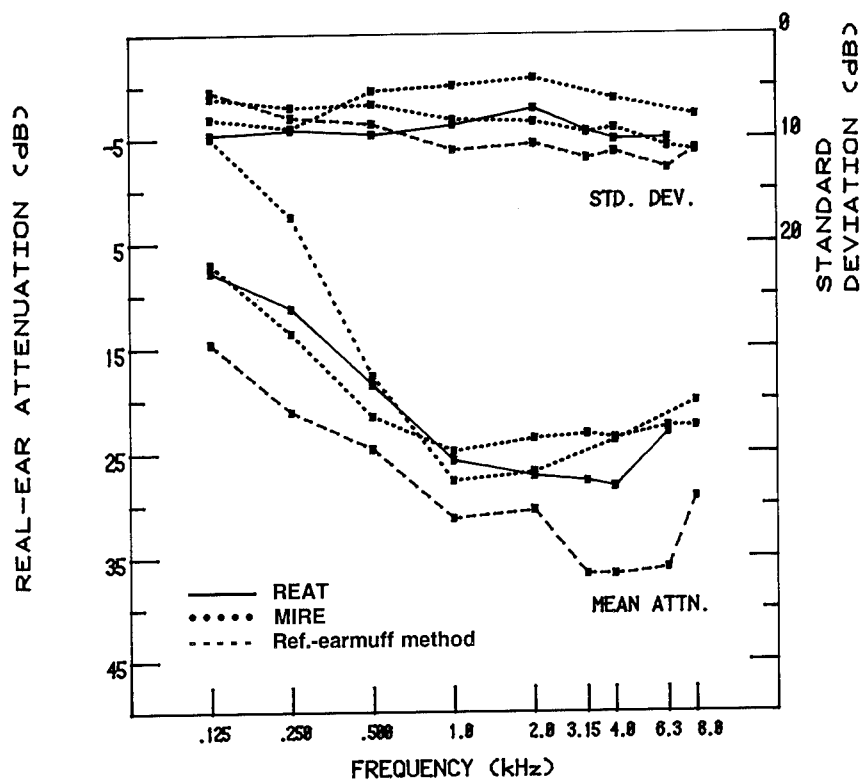


Figure 29-7 Real-world performance of the MSA Mark IV earmuff (four studies, 89 subjects).

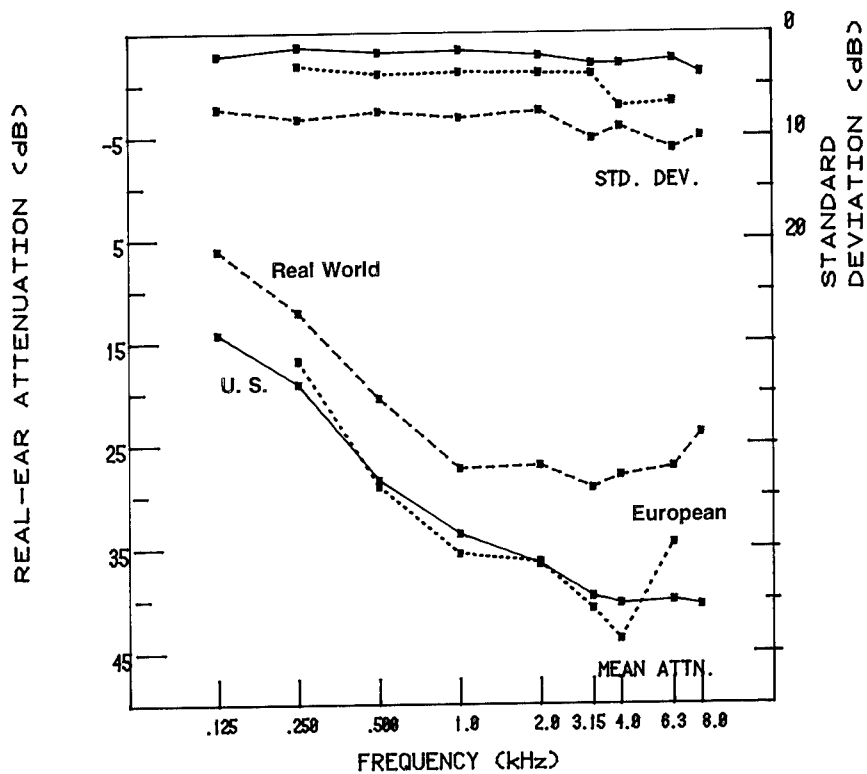


Figure 29-8 MSA Mark IV earmuff: average real-world attenuation compared to manufacturer's US test data and European laboratory results.

Following are specific observations about the data:

(1) Based upon real-world data, the lowest attenuating earplug among devices thus far tested, is the EP100. This is due to low mean attenuation values and high variability. Four of the five field studies agree rather closely (within 7 dB up through 2 kHz) (Figures 29-1, 29-2).

(2) The P.O.P. earplug exhibits a very tight range in mean attenuation values and SDs across field studies. The spread in data is about what would be expected from a typical interlaboratory as opposed to an interworkplace study (Figures 29-3, 29-4).

(3) The E-A-R/Decidamp earplug provides potentially high degrees of protection, but also a wide range of attenuation and SD values across 12 separate studies. The variability is probably due to the fact that foam plugs, although they seal the ear well regardless of insertion depth, can provide dramatically differing values of attenuation depending upon the depth of insertion. Insertion depth of foam earplugs is a parameter that is heavily influenced by subjects' training and motivation to properly use the product, and also may be affected by the amount of noise reduction the wearers require or desire. (Figures 29-5, 29-6).

(4) The earmuff data include measurements from three different types of studies. The fact that the data from the reference-earmuff method are the highest shown, may be due to the way in which those real-world employees actually wore their earmuffs, or may be experimental artifact as discussed earlier. The averaged earmuff results shown in Figure 29-8 are representative of those found for other earmuffs, with the exception of the real-world SDs that tend to be high for this particular product. The differences between US and European mean attenuation values are insignificant, but the SDs are higher for both the European and the real-world data than for the US results (Figures 29-7, 29-8).

(5) Figure 29-9 provides a comparison of standard headband earmuffs to hard hat at-

tached earmuffs. Despite the dissimilarity in the way the two types of earmuffs interface to the head, no practical differences were found in their real-world performance, that is, mean attenuation values were within 2.6 dB, and SDs within 1.2 dB at all frequencies.

Real-world data and US test data were compared for three earplugs and one earmuff for which there were sufficient samples for analysis. The mean real-world attenuation values were found to be statistically significantly smaller, and the associated SDs significantly larger, than for US laboratory data. There was more degradation in earplug than in earmuff performance, as would be anticipated due to the greater difficulty in fitting and inserting earplugs than earmuffs, but the differences were unique to the HPD tested. A similar analysis was not performed for the European laboratory data. However, as has been previously observed,³² they appear to provide a closer approximation to real-world values than do the US data.

REAT Versus MIRE

Figure 29-10 depicts the real-world data for more than 16 models of earmuffs separated into nine REAT (501 subjects) and four MIRE (315 subjects) studies. Four interesting observations are apparent:

(1) Over the middle frequencies from 500 to 2000 Hz, where both methods are devoid of experimental artifact, the mean attenuation results of the two procedures are in nearly exact agreement, despite the wide diversity of samples and studies that are combined to produce the averaged results. No evidence is seen of any aberration due to learning effects, which would have caused the REAT values to exceed the MIRE data.

(2) As is well-documented in the literature, REAT yields spuriously high values of attenuation at the low frequencies due to physiological noise masking the thresholds in the occluded condition, and hence inappropriately increasing the occluded/open threshold shift.²⁷ At such frequencies an ob-

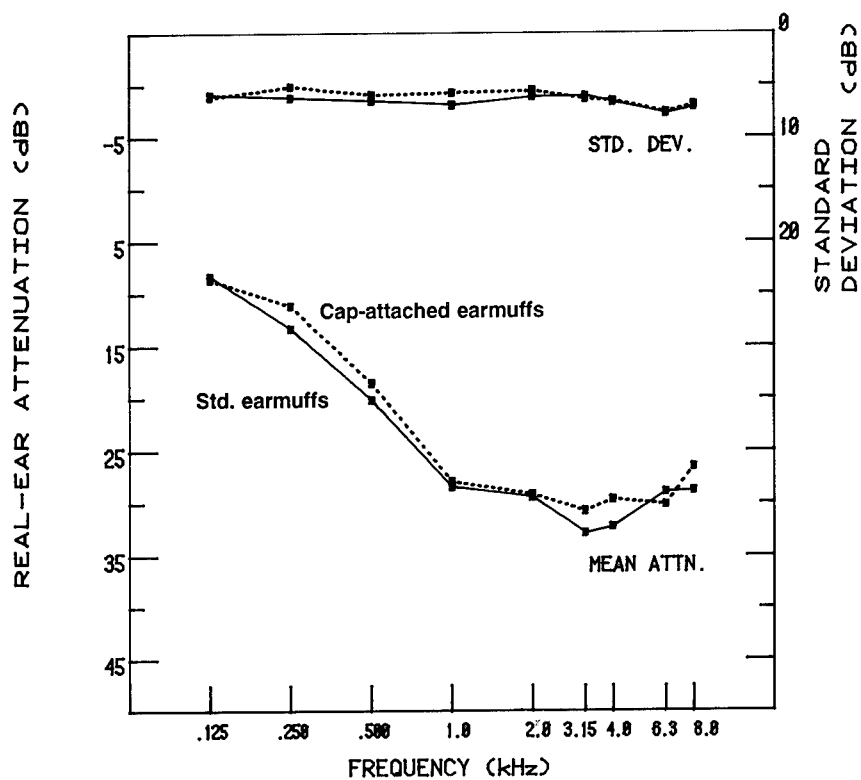


Figure 29-9 Comparison of standard earmuffs (eight studies, 324 subjects) to cap-attached earmuffs (four studies, 177 subjects) using real-world REAT data.

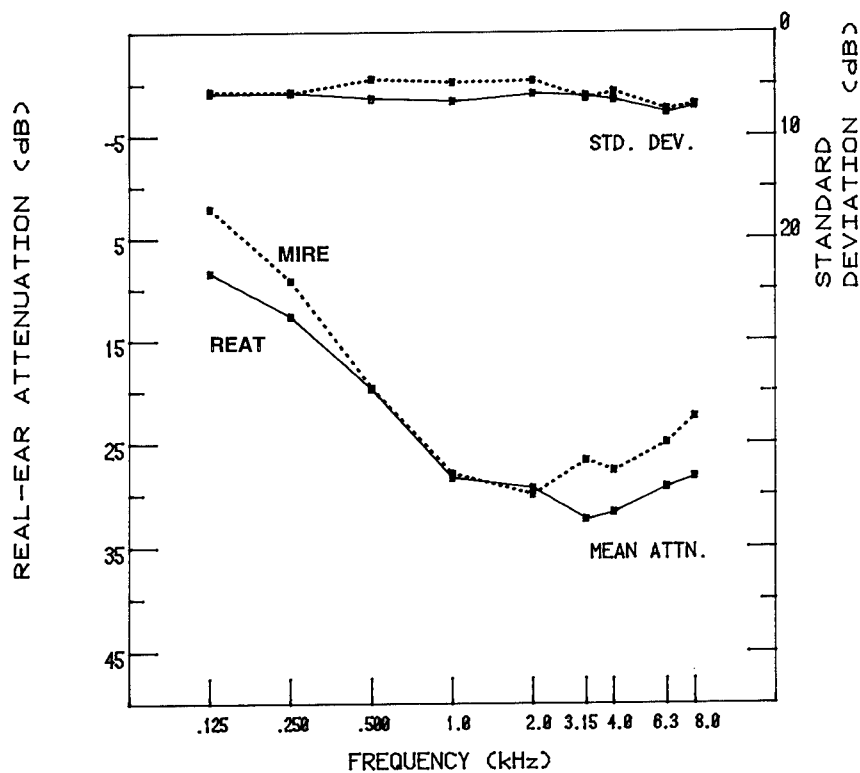


Figure 29-10 Comparison of real-world earmuff attenuation measured using REAT (nine studies, 501 subjects) and MIRE (four studies, 315 subjects) procedures.

jective measurement such as MIRE is more appropriate. The REAT/MIRE disparity in Figure 29-10 is seen to be from 6 to 3 dB at 125 and 250 Hz, respectively, in agreement with previously reported laboratory results.

(3) As discussed earlier, field implementation of the MIRE procedure is typically based upon NR instead of IL measurements, which leads to underestimates of attenuation above 2 kHz. This can be clearly noted in Figure 29-10. Therefore, REAT data, which are devoid of high-frequency artifact, provide the better assessment of attenuation at high frequencies.

(4) Concern is sometimes expressed that real-world REAT studies yield excessively high values of SD because subjects are not adequately trained in taking threshold audiograms, and thus their threshold variability contaminates results. If so, one would expect that an objective measurement such as MIRE, which does not include a threshold-variability component, would indicate lower SDs, and thus provide SD estimates more representative of the true variability in fit of the HPDs between subjects. This was not the case. At four of the seven test frequencies the SDs are essentially identical for both methods; from 500 to 2 kHz where differences exist, they amount to less than 2 dB.

Discussion

To more easily compare device types and gain a perspective of the attenuation attainable in the real world, data for three-flanged pre-molded earplugs, custom-molded earplugs, sheathed fiberglass earplugs, vinyl foam earplugs, and earmuffs, are compared in Figure 29-11. Foam earplugs provide the highest attenuation at 125 and 250 Hz and above 2 kHz, and earmuffs the most attenuation in the middle-frequency range, from 500 to 1000 Hz. In addition to the octave-band data, the NRR_{84} and the HML values²⁴ were also computed with a 1 SD correction and listed below the graph. They tell a similar story.

Note that the earmuffs show the smallest SDs at all frequencies, again confirming the

greater ease with which they can fit, or be fitted by, a wide-ranging group of people.

The NRRs of the five device types were tested by a one-way analysis of variance, and found to have a significant device effect at $p < 0.001$. However, subsequent tests demonstrated that the custom-molded, fiberglass, and three-flanged groups were not significantly different at the $p < 0.05$ level, and that likewise the differences were not significant between the foam earplug and earmuff categories. Thus in terms of overall protection, the real-world data suggest that it is not possible to make fine distinctions between types of hearing protectors. To a first approximation only two categories can be distinguished: one consisting of the higher attenuation devices of foam earplugs and earmuffs, and the other consisting of lower attenuation devices comprised of the remaining principal types of (nonfoam) earplugs.

As an additional summary of the real-world data, Figure 29-12 provides an overview in terms of the field NRR_{84} s versus the manufacturers' published laboratory NRR_{98} s. The same trends emerge as were apparent in Figure 29-11. Measured as a percentage of the laboratory-rated attenuation, the field NRRs for earplugs yield only about 25% of the labeled values, and for earmuffs about 60%. It is especially clear that the American laboratory data not only provide a poor indication of the absolute values of field performance, *but of the rank ordering of those values as well*. This means that no single correction factor can be applied to existing laboratory data to estimate field performance. This is also demonstrated by the data in the last column of Table 29-1 that lists the real-world NRR_{84} as a percentage of the labeled NRR.

Especially misleading is the fact that the laboratory data would suggest that in general, earplugs provide the highest overall protection whereas, with the exception of foam earplugs, the reverse is true under field conditions.

Although the current report is intended primarily to provide a real-world data base for use in future research, it is instructive to discuss potential reasons for the divergence be-

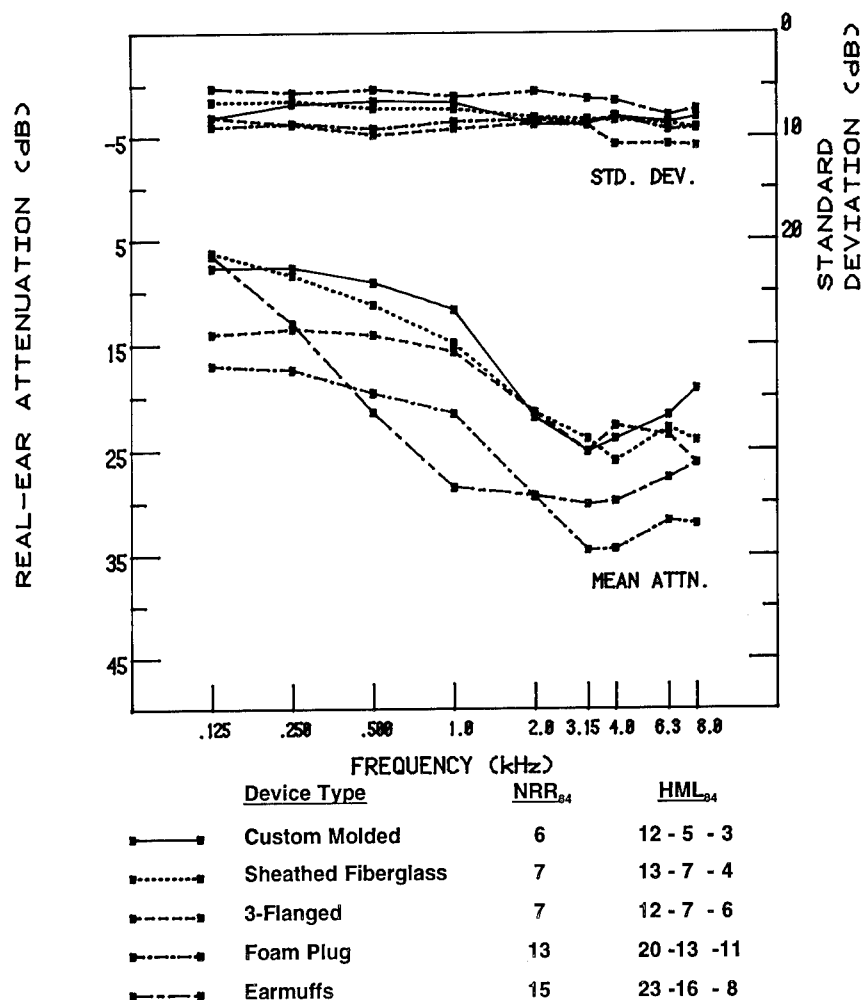


Figure 29-11 Summary of real-world data for hearing protectors separated into five categories.

tween laboratory data (primarily those of US origin) and field performance, most substantially for earplugs, but to a noticeable extent for earmuffs as well. The problem of predicting real-world performance has been extensively studied by S12/WG11 and has been the subject of research presentations as well as work in progress on a draft standard.³⁶

A portion of the lab/real-world divergence is due to less than desirable quality in real-world hearing conservation practices in areas of fitting and training of HPD users, enforcement of proper HPD utilization, education and motivation of the work force, and program management. And the fact must be

considered that user fitting of HPDs in the real-world is strongly affected by comfort, convenience, and interference with communications, whereas in the laboratory environment these parameters are considerably less important than attenuation.

Much of the divergence between laboratory and real-world data is also attributable to inappropriate laboratory practices and consequent unrealistic test results. It is just those practices, in the areas of subject selection, fitting, and training, as well as experimenter involvement and consistency across facilities, that are being addressed by S12/WG11. Based upon results of a four-facility interlaboratory study

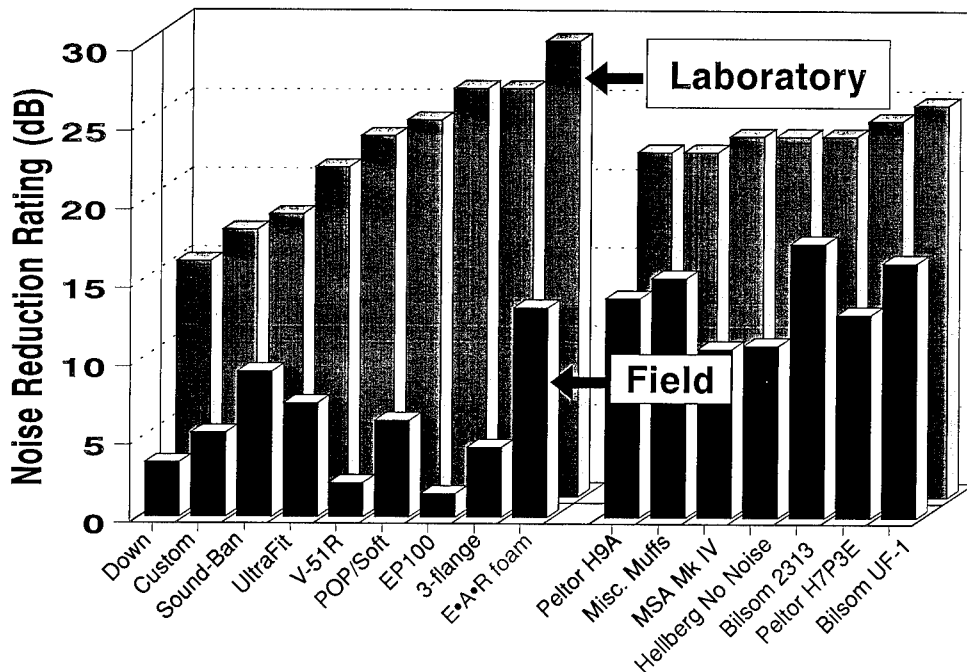


Figure 29-12 Comparison of NRRs published in North America (labeled values based upon laboratory tests), to real-world "field" attenuation results derived from 22 separate studies.

conducted under the auspices of the working group, there is optimism that a solution can be devised.³⁶

Conclusions

Although the data base has grown substantially larger since the appearance of the earliest studies and summary reports,²³ the conclusions remain the same: real-world performance of HPDs, especially earplugs, demonstrates less attenuation and greater variability than currently standardized laboratory tests would predict. Measured in terms of the overall protection achieved by 84% of the workforce, earplug attenuation varies from a low of 1 dB for one type of premolded earplug to a high of 13 dB for foam earplugs, and about 11–17 dB for earmuffs.

Because field data are normally examined in terms of a value achieved by 84% of the users, the attenuation values appear quite low. However, field SDs are normally around 8–10 dB, and thus when the protection values are

increased by 1 SD to estimate a mean value instead of an 84th percentile value, considerably larger amounts of attenuation are predicted. The selection of the statistical adjustment to include in the computation depends upon the goals of the specifier.

Field attenuation values are low enough that in many actual environments, even when only 10 dB of attenuation is required, it is questionable whether certain HPDs can provide the degree of protection needed for the majority of the workforce. Such findings may appear incredible to some observers, but the magnitude of the results is qualitatively supported by analyses of audiometric data from existing hearing conservation programs, and by real-world studies of temporary threshold shift.³⁷

On a global basis there is no question that the existing group of 22 studies provides a clear indication of field performance, but additional data are required if specific guidance is to be developed for a wide variety of individual devices. HPDs that are in particular need

of additional field studies are the semi-insert/semiaural types of hearing protectors as well as dual hearing protection, that is muffs and plugs worn in combination, the latter category for which (to the authors' knowledge) no published data on real-world attenuation are yet available.

Current research has demonstrated that a good estimate of the real-world attenuation achieved in the better programs can be obtained by testing *totally naive HPD users in a laboratory protocol with absolutely no individual training by the experimenter*.³⁶ When tested under those conditions, the attenuation of HPDs still equals or exceeds average real-world data of the type shown here. The fact that subjects completely untrained in the use of HPDs obtain more attenuation than occupationally exposed workers who would have been expected to be trained and motivated and to have benefitted from many months of practice in using their HPDs, is truly amazing! It suggests that today's typical, or even above-average hearing conservation programs, are ineffective in fully motivating and training employees to consistently and properly wear their HPDs.

Regardless of these issues and the research that is still needed to better define field performance possibilities, use of HPDs remains key to the prevention of occupational noise-induced hearing loss. If only hearing protection devices were worn properly and consistently, such causes of hearing loss would cease to exist.

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Chapter 30

Distributions of Hearing Threshold Levels in Populations Exposed to Noise

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The aim of this chapter is to give a broad overview of knowledge on the effects primarily of occupational noise on hearing threshold levels (HTLs). The first part of the chapter concerns work published over the past 25 years, mainly based on studies conducted at least 20 years ago of hearing impairment in people exposed to very high levels of occupational noise. The second part of the chapter concerns more recent studies of hearing in the general population that included people with experience of occupational noise exposure as well as people without material noise exposure. These two different approaches each have advantages and disadvantages, and provide data on the effects of noise on hearing that are complementary.

Despite the high prevalence of hazardous levels of occupational noise exposure in industrialized society, systematic examination of its long-term effects on the human auditory system is beset by considerable difficulties. Deliberate experimental manipulation of parameters of noise exposure (dose) in order to assess their effect on human hearing thresholds is not ethically justifiable in view of the permanent damage that is generally caused. Even short-term noise exposures that cause relatively minor temporary shifts of hearing threshold may lead to irreversible damage that is not evident in the crude measure of hearing sensitivity given by the pure-tone audiogram, and hence human studies of temporary threshold shift (TTS) that were common two or three decades ago are avoided nowa-

days. These constraints entail investigations of noise-induced hearing loss (NIHL) that are primarily observational, relying on noise exposures as they occur in the workplace rather than under experimental control. Furthermore, such studies are primarily retrospective rather than prospective. To be generally applicable, a good prospective study would have to observe over a long time, for example, from the start of employment until after retirement. Observation of the attendant loss of hearing without intervention to prevent further damage would also be unjustifiable from an ethical standpoint.

The above restrictions lead to substantial difficulties of interpretation of the available data for several reasons. First, estimating noise exposure levels and durations retrospectively is subject to both random and systematic error components. Although current measurements can give an estimate of historical noise levels, processes and working practices change with time and the degree of representativeness of current levels is difficult to assess. Furthermore, noise exposure patterns vary, and the extent of exposure will depend on the location of the individual in relation to such variations. Second, lack of control of noise exposure parameters makes it difficult to separate out the effects of noise exposure characteristics such as level, frequency content, and temporal pattern. Third, most noise exposures are such that it takes several years before the effects of noise damage are clearly evident against the background of intersub-

ject variation, and hence there are inevitably effects of aging to be taken into account, entailing comparison with nonexposed controls. Fourth, there is no possibility to allocate subjects randomly to exposed or nonexposed (control) groups, and hence great caution is required in interpreting differences between exposed and control groups. Such differences could be due to confounding factors often neither measured nor reported. Moreover, studies are often restricted to specific groups of individuals, rather than being based on random samples of the general population.

A fifth difficulty might be added to the above list. Improved attitudes to and regulation of occupational noise exposure have reduced the numbers of people exposed to high levels of noise. This restricts the availability of new data that show clear effects of noise exposure. Hence, analysis is based primarily on studies conducted some years ago when standards of measurement of noise levels and of hearing thresholds were less well developed than at the present time. In some studies, HTLs were not referenced to established standards and analysis must resort to the biological benchmarks provided by small groups of control subjects included in the studies. This leads to an additional source of error. A consequence of the above limitations is that most analyses of the effects of noise have made at least some assumptions about the relationships between the physical characteristics of the noise and their effect on hearing, about the effects of age, and about the way in which noise and age effects interact. Many of these assumptions have not been evaluated to any great degree of accuracy, and any interpretation of the analyses must bear in mind the consequences of any deviation from the assumptions. A few of the common assumptions are outlined below.

Characteristics of Noise Exposure

It is almost universally assumed that allowance for different noise frequency spectra can be made by expressing the noise level in terms of the A-weighted sound pressure level

(SPL). This proposition was evaluated by Burns and Robinson¹ and there has been no general opposition to this approach. In fact, the use of A-weighted SPLs is endorsed in many national and international standards and regulations concerning risk of hearing damage due to noise (e.g. ISO 1999²). By contrast, there has been considerable contention over methods used to allow for temporal fluctuations in noise level. Temporal fluctuations may occur over seconds, minutes, hours, days, and even years. Fluctuations may be intrinsic to the noise itself, or may be due to work patterns. For example, the individual may be exposed to the noise for four half-hour periods on each of 2 d/wk, making a total of 4 h/wk. Furthermore, different individuals will be exposed for different overall lengths of time, and some may be exposed to one type of noise pattern for several years followed by a different pattern for a further period. In fact, exposure to the same steady level for each working day throughout a working lifetime is the exception rather than the rule. Therefore, to perform a tractable analysis of real-world data, a method must be used to amalgamate data from individuals with somewhat heterogeneous patterns of noise exposure. One method that has been used extensively is to apply the "equal-energy" principle. This states that noise exposures with an equal (A-weighted) energy content have an equal effect on hearing thresholds, regardless of the distribution of sound energy over time. This principle can be applied over relatively short periods to obtain an equivalent daily exposure, over longer periods to obtain equivalent annual exposures, or over an entire working lifetime to equate noise exposures with different levels that extend for different numbers of years. Although there is no theoretical basis to the equal-energy principle, there is a substantial body of evidence supporting its use to obtain equivalent daily exposures.³ However, its use relating to the long-term accumulation of NIHL is equivocal and different analyses have taken different approaches to pooling data that consist of various noise exposure patterns.

Effects of Age on Hearing

The effects of age on hearing threshold levels have been widely studied and have reached the status of an international standard, at least for that fraction of the population referred to as "highly screened" (Database A of ISO 19992). Implicit in this definition is the notion that screening acts to sift out subjects with identifiable ear pathology who would otherwise weight the results toward poorer hearing. When such screening is omitted in large population surveys, a different data set arises termed "typical population" data or Database B.⁴ Such typical population data have a median value that is approximately 10 dB less sensitive than highly screened data, the difference being approximately constant with increasing age. With such a large difference between highly screened and typical population data, the methods used to allow for the effects of age can have a profound effect on the interpretation of hearing thresholds of noise-exposed populations.

Although it is generally assumed that the differences in hearing thresholds between highly screened and typical population data sets is a direct result of the screening, there is evidence to suggest that even the most rigorous otological screening cannot account for a difference of more than a few decibels at the median, at least in younger adults.^{5,6} Substantial differences may arise out of other characteristics of the subjects actually sampled, and further smaller differences can arise from the audiometric procedures used. Hence, proper matching of controls to noise-exposed groups involves more than just otological screening. These issues will be discussed further.

Interaction Between Noise Exposure and Age

There have been a variety of approaches to dealing with the combined effects of noise and age. At one extreme, the hearing losses due to age and noise are added (in decibels) to give the overall hearing loss.¹ At the other extreme, they are converted into their underlying units of pressure, added, and converted

back to decibels.⁷ The difference between these two approaches is best appreciated by an example: if the effect of noise exposure on hearing threshold level is 30 dB and that due to age is 20 dB, then the former approach gives a combined hearing loss of 50 dB, whereas the latter approach gives a combined hearing loss of 30.4 dB. In between these extremes, ISO 1999 assumes that the two components are additive (in decibels) for mild hearing losses, but that the addition becomes increasingly incomplete as their magnitudes increase.²

One of the reasons for the wide divergence between these assumptions is that the available data give very limited leverage. In general, individuals exposed to noise are exposed to a cumulative noise dose that increases steadily over many years. Inevitably they have become older over the same period of time and hence age and cumulative noise exposure are inextricably confounded. The extent to which different theoretical models fit the data is rarely tested rigorously. Although competing models may be compared, little attention is given to whether the assumptions behind the model are satisfied. Even those studies that have attempted to distinguish the relative effects of age, noise, and their interaction⁸ are rather insensitive to differences in the additive model.

Existing Experimental Data

NIHL

Sets of data used to examine the effects of noise exposure in humans generally come from two sources. First, the data from noise-exposed subjects take the form of ages and HTLs measured after an extended period of noise exposure. The noise exposure is usually described in terms of an A-weighted noise level in decibels and a duration in years. This level may have in fact been relatively constant at that level, or it may have fluctuated somewhat giving an equivalent level equal to the quoted noise level. Typically, the equivalent level is calculated on an equal-energy basis. Second, there are data from controls who have not been exposed to hazardous

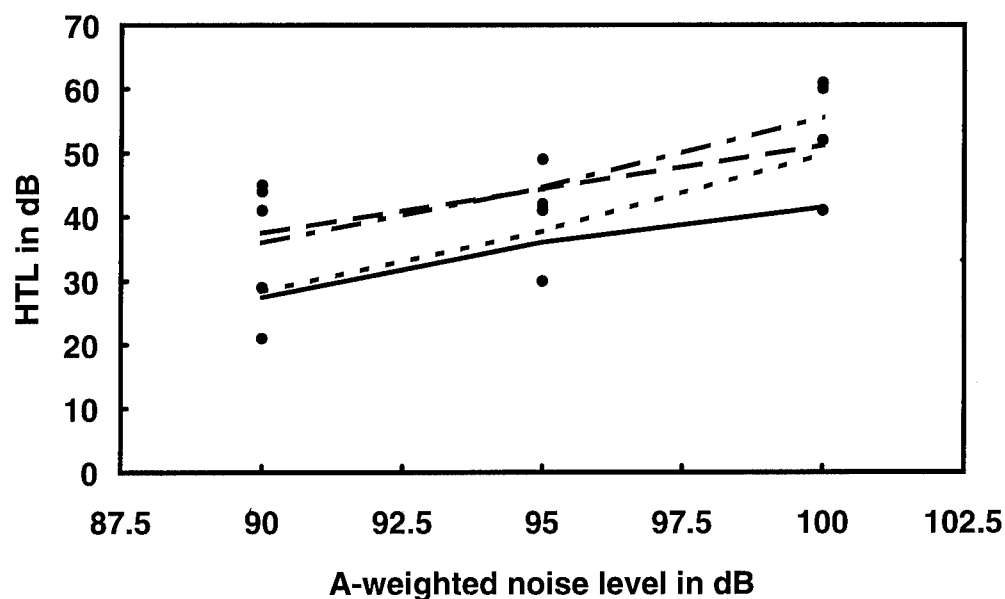


Figure 30-1 Comparison of the outcomes of various studies, data bases, and formulas concerning median hearing threshold levels (HTL) at 4 kHz in men exposed to occupational noise for 30 years at A-weighted levels of 90, 95, and 100 dB(A). (●) All studies represented in figure 4.13 of Robinson.⁸ (—) NPL Tables.¹⁰ (---) HSE.⁸ (-.-) ISO 1999² with Database A.¹³ (....) ISO 1999² with Database B (IHR screened²⁵).

levels of noise, which are in the form of ages and HTLs. It is assumed here that all subjects are of the same sex and this discussion will focus primarily on males, for whom the prevalence of occupational noise exposure is higher.⁹ The data sets also usually contain statements about the screening that has been conducted to exclude subjects with ear pathology other than that due to noise exposure or due to the effects of age. At this stage, the raw data are uncontaminated by the various assumptions that are made when interpreting the data, other than any assumptions used to derive the equivalent (annual) noise level. It is instructive to examine the raw data before any further processing. For the sake of simplicity, examples will be restricted to the medians from certain studies, and the medians from secondary analytic studies or standards that are based on similar primary studies. It is well known that NIHL is typically greatest at around 4 kHz and the examples are restricted to that frequency in order to emphasize the difference between noise-exposed subjects and controls. Figure 30-1 shows median hear-

ing threshold levels at 4 kHz in males aged 50 years after exposure to noise for 30 years at three different levels: 90, 95, and 100 dB(A). The symbols are from selected studies and the lines are from secondary studies or standards. The latter data sets are presented in terms of formulas that have been constructed to fit primary data sets. Not surprisingly, the formulas tend to follow the symbols in Figure 30-1 as they are derived from the same or similar data sets. However, the most striking feature of Figure 30-1 is the large range of HTLs encompassed by the various medians. The apparent increment in HTL resulting from an increase in noise level from 90 to 100 dB(A) is smaller than the range of values at any of the noise levels. Stated in another way, there is marked overlap between the values at each of the noise levels such that the largest value corresponding to a noise level of 90 dB(A) is substantially larger than the smallest value corresponding to a noise level of 100 dB(A). It should be emphasized that these data are medians from sizeable studies and are not case studies of individual subjects. In summary,

Figure 30-1 shows that the magnitude of the effect of noise even at 4 kHz is small compared with the variation between studies. These observations indicate that any inferences that can be drawn about the effects of noise are of a broad statistical nature relating to populations. They cannot be applied directly to individuals with any semblance of accuracy.

The formulas corresponding to the four lines in Figure 30-1 are based on three methods for estimating the effect of noise on HTLs, one of which has been used with two sets of data to represent the effects of age. Brief descriptions of the three methods follow.

National Physical Laboratory

The NPL Tables¹⁰ are based on a study of NIHL conducted by Burns and Robinson.¹ They are generated by a formula that predicts HTLs and their distribution as a function of noise immission level (NIL), where NIL is a cumulative measure of noise exposure based on the equal-energy principle. Strictly, the NPL Tables estimate the noise-induced component of HTL, which may then be added to an age-related component that is also tabulated, based on the data of Hinchcliffe.¹¹ Implicit in this method, therefore, is the assumption that age and noise components add arithmetically in decibels. However, given that the raw data that gave rise to the formula consisted of noise exposures that were steady and started at approximately the age of 20 years for all subjects, using the NPL Tables to give the line in Figure 30-1 effectively reverses the process of separating out age and noise effects, and should therefore regenerate the raw data independent of any assumptions about additivity.

Health and Safety Executive

This report produced by Robinson⁸ synthesizes a large data base of 17 existing studies of NIHL to provide a formula for predicting HTLs as a function of noise level and duration. The data did not support the concept of NIL for uniquely expressing cumulative noise exposure on the equal-energy principle. Nor did

the data support the simple addition of age and noise components in decibels. Instead, the combined effect of age and noise was found to be somewhat less than the arithmetic sum of the separate threshold shifts, the reduction increasing according to the product of the two threshold shifts to an extent that depended on audiometric frequency. This idea was encapsulated in the formula shown in Eq (30-1).

$$H = A + N - A \times N/k \quad \text{Eqn (30-1)}$$

where H is the combined HTL, A is the component of HTL due to age, N is the component of HTL due to noise, and k is a constant depending on audiometric frequency.

The constant k was found to be smallest for the highest audiometric frequency considered (4 kHz), and to increase with reducing audiometric frequency. Hence, the deviation from simple addition was greatest at 4 kHz. It should be noted that this and most other such models have been primarily deterministic, and have not attempted to model the error distribution.

The noise term in Eqn (30-1) is related to the noise level and duration such that N increases with an exponentially decelerating time course toward a limiting value, N_{\max} , the latter value depending linearly on the A-weighted noise level at each audiometric frequency. Hence, for a given frequency and noise duration, N depends linearly on noise level. The age term in Eq (30-1) as originally formulated was a mixture of highly screened and typical population data, according to the balance of screened and unscreened studies contributing to the analysis.

ISO 1999

This standard provides a formula for estimating distributions of HTLs in noise-exposed populations based on studies by Burns and Robinson¹ and Passchier-Vermeer,¹² the latter being a review of earlier studies. It uses Eq (30-1) to allow for incomplete addition of age and noise effects, but with k fixed arbitrarily at 120. The origins of this value are unclear, but it

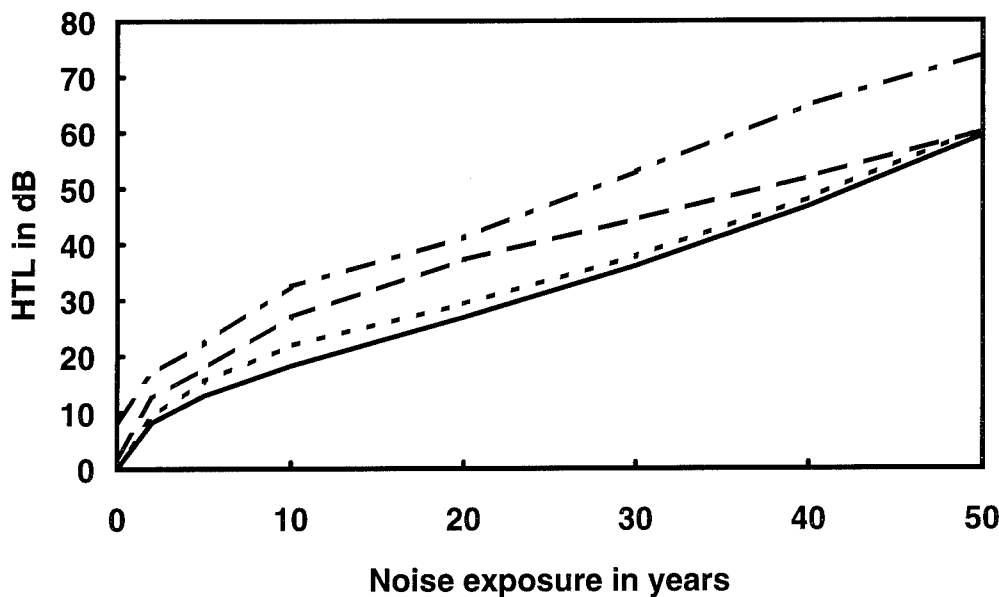


Figure 30-2 Growth of median hearing threshold level (HTL) at 4 kHz as a function of duration of exposure to noise at 95 dB(A), according to four formulas. Noise exposure is hypothesized to start at the age of 20 years. Points calculated at durations of 0, 2, 5, 10, 20, 30, 40, and 50 years, and interpolated linearly between. (—) NPL Tables.¹⁰ (---) HSE.⁸ (---) ISO 1999² with Database A.¹³ (-.-) ISO 1999² with Database B (IHR unscreened).¹⁴

has the effect of ensuring that H cannot exceed 120 dB.

The noise term in ISO 1999 increases logarithmically with duration for a given noise level. The effect of noise level is related to the square of the amount by which the noise level exceeds a cutoff level. This cutoff level varies according to audiometric frequency between 93 dB (0.5 kHz) and 75 dB (4 kHz). The age term is not defined exclusively in ISO 1999, but is left to the choice of the user, depending on the characteristics of the population in question. Two categories of age-related data base are described: Database A obtained from otologically normal subjects, such as the "highly screened" values given by ISO 7029,¹³ and Database B which relates to unscreened populations. ISO 1999 gives an example of Database B as an appendix but recommends that the user should provide locally applicable values. The two lines in Figure 30-1 derived from ISO 1999 differ only in the choice of data base to represent the effects of age. The lower line uses Database A (ISO 7029), and the up-

per line uses unscreened population data from the United Kingdom.¹⁴

The differences between the above formulations can be appreciated by comparing the HTLs predicted. Figure 30-1 already compared predictions at 4 kHz (median) for exposures starting at age 20 and finishing at age 50. Figure 30-2 compares the growth of HTL at 4 kHz over time for a fixed noise level of 95 dB (A) using the same formulas and age-related data sets. Three of the four lines use similar age-related data sets which leads to them starting at similar values before the beginning of exposure hypothesized to be at the age of 20 years. They also converge toward similar values after 50 years of exposure (age 70) as the age component dominates the overall HTL. Not surprisingly, the ISO 1999/ ISO 7029 line and the NPL line are very similar, given their close connection with the Burns and Robinson¹ data set. The main difference between these three curves is the rate at which the noise-induced component develops over the first 20 years of exposure (when the age-

related component is small), and the extent to which the noise-induced component tends to saturate. In particular, the Health and Safety Executive (HSE) curve tends to accelerate more rapidly and shows stronger saturation than the other two curves. The fourth curve, based on ISO 1999 and unscreened population data obtained by the Institute of Hearing Research in the UK (IHR-U) is shifted toward larger HTLs at all exposure durations. This is because the IHR-U data are systematically elevated compared to ISO 7029. It should be noted that the use of ISO 1999 with such age-related data, although recommended by the standard, is untested against measured HTLs in noise-exposed populations.

Age-Related Hearing Loss

Values of age-related hearing loss have already been used to calculate the examples in Figures 30-1 and 30-2. Several data sets exist from which to draw age-related HTLs, one of which has achieved the status of the international standard, ISO 7029 (1984). Figure 30-3

compares the median values of ISO 7029 at 4 kHz as a function of age with four other data sets. All five lines in the figure relate to male subjects. The line labeled NPL is the set of age corrections in the NPL Tables. The line labeled TP is the typical population data set of Robinson,⁴ based on two very large population studies in the United States. The remaining two lines are from more recent population studies in the United Kingdom conducted by the Institute of Hearing Research. Those studies are outlined later in this chapter. One line (IHR-U) is for unscreened samples and the other (IHR-S) is for samples screened for absence of noise exposure, and absence of middle ear disorder based on the air-bone gap measured audiometrically.

Figure 30-3 contains some striking similarities. First, the unscreened data (TP and IHR-U) are remarkably similar, given their provenance from independent studies conducted in different continents approximately 2 decades apart. Second, the two large screened data sets (ISO 7029 and IHR-S) are remarkably similar for subjects aged 45 years

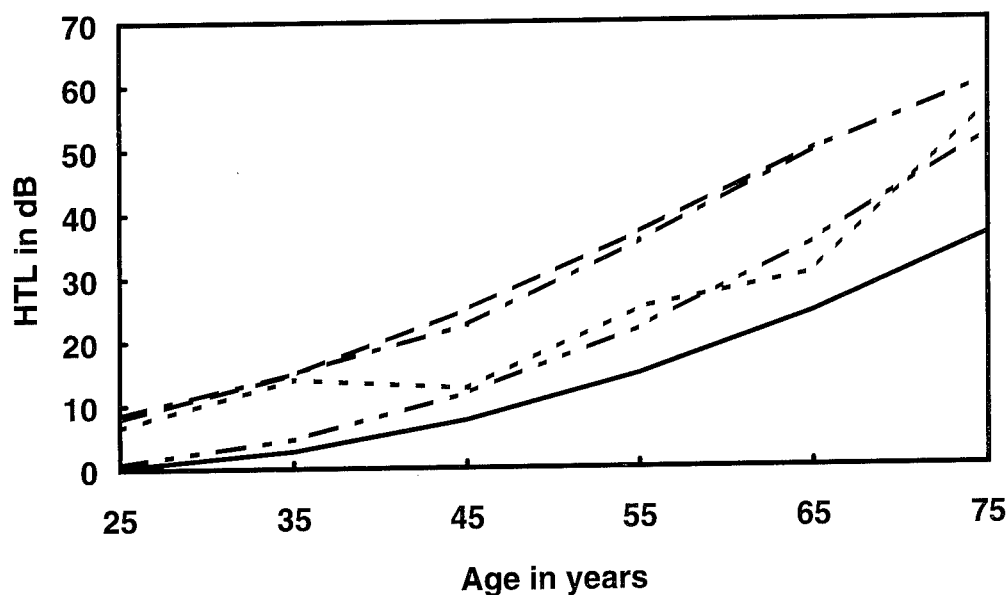


Figure 30-3 Comparison of median age-related hearing loss at 4 kHz, according to five data bases and formulas. Points calculated at 10 year intervals and interpolated linearly between. (—) NPL Tables.¹⁰ (---) ISO 7029.¹³ (···) Robinson.⁴ (— · —) IHR-U, unscreened.¹⁴ (- - -) IHR-S, screened.²⁵ Note, IHR data are unsmoothed estimates.

and above. The two curves deviate quite strongly, however, for younger subjects with IHR-S approaching the unscreened data sets. The lack of differentiation of screened and unscreened medians in the IHR data sets arises out of the low prevalence of middle ear disorder in younger subjects. The NPL age corrections, as tabulated, indicate relatively good hearing for older subjects. They are based on a study by Hinchcliffe¹¹ of 318 subjects in a rural districts of Scotland. However, the original Hinchcliffe data were reported as HTLs relative to a control group. It is tempting to observe that shifting the NPL data upward by 10 dB would bring them closely into alignment with the IHR-S values.

Despite the specific similarities within Figure 30-3, the main impression is the wide range of the estimates from different sources. Taking a single age value of 50 years (for comparison with Figure 30-1), the range of estimates for the median HTL at 4 kHz is from 10 to 30 dB. By comparison, the ranges of estimates for the median HTL at 4 kHz in noise-exposed populations is equally wide and overlaps the range for nonexposed populations for noise levels of 90 and 95 dB(A). It is not until the noise exposure reaches 100 dB(A) daily for 30 years that the ranges no longer overlap. This comparison has deliberately been drawn at 4 kHz where the effects of noise are generally greatest, and at an age of 50 years before the age-related component tends to swamp the noise component. At younger ages the total noise exposure is limited due to shorter duration, and it is therefore more difficult to demonstrate an effect of noise.

The above comparison is intended to illustrate a number of important considerations that should be borne in mind when attempting to define the effects of noise on hearing. First, it is often difficult to demonstrate *any* systematic effect of noise exposure due to the overlapping distributions of exposed and control subjects. Second, the nature of the relationship is generally most readily observed at 4 kHz; at other frequencies it is not always possible to define the relationship clearly and it is common practice to *assume* similarities with the relationship at 4 kHz. Third, given

the variations among nonexposed populations, differences between exposed and control subjects can depend strongly on the choice of controls. Fourth, small differences in the formulation of standards for predicting NIHL are probably unsupportable on the basis of experimental data; the variability inherent in existing data sets and the possibilities of bias in control groups entails that only broad conclusions can be drawn. In addition to the above considerations, it is desirable that models of NIHLs should have a justifiable theoretical basis in addition to fitting empirical data.

It is clear from comparison of Figures 30-1 and 30-3 that very different impressions of the effect of noise on hearing could be obtained, depending on the choice of data sets for noise-exposed and control subjects. By deliberately choosing the noise-exposed data set with the highest HTLs and the control data set with the lowest HTLs, the effect of noise would appear to be rather large, reaching approximately 38 dB for a noise level of 95 dB(A) and an age of 50 years. By contrast, deliberately choosing the noise-exposed data set with the lowest HTLs and the control data set with the highest HTLs, it would appear that the same conditions of noise exposure had no effect. (These uncertainties are exacerbated when considering the effects of lesser noise exposures, a topic of great interest at the present time in relation to national and international regulation of workplace noise.) Thus, it is of overriding importance to ensure that the control data set is chosen carefully to reflect as accurately possible the hearing status of the noise-exposed group that would have existed if the latter had not been exposed to noise. One way of doing this is to use an epidemiological survey approach, using identical sampling methods for both groups in the same population. In practice, this entails drawing a random sample from a suitable population and assigning subjects to noise-exposed and control groups based on their previous noise exposure history. One such epidemiological survey is the UK National Study of Hearing (NSH), relevant parts of which are described below. The data from the NSH provide infor-

mation on the effects of noise on hearing that are complementary to those outlined above.

UK National Study of Hearing

Background

During the early 1980s, the Medical Research Council's Institute of Hearing Research conducted a large population study of hearing in adults in the UK. Although not specifically designed to investigate NIHL in detail, results from the NSH throw some light on the issues discussed above. The main strength of the NSH in this context is that it employed formal random sampling of the general population. This allowed demographic trends to be determined. In particular, random sampling reduces the possibility of bias between groups of noise-exposed and noise-free subjects that are compared, and the availability of relevant demographic variables allows the groups to be either matched on demographic characteristics, or for the influence of such variables to be accounted for statistically. One particular variable that is important in this respect is socioeconomic status or social class, determined according to the Registrar General's classification of occupations in the NSH, and dichotomized in terms of occupational group (manual or nonmanual occupation) in the present analysis. Manual occupation is consistently associated with poorer hearing than nonmanual occupation,¹⁵ even in noise-free groups. Given the higher prevalence of noise exposure in manual occupations, lack of control for occupational group can lead to erroneous conclusions when comparing noise-exposed and noise-free groups.

We have previously addressed the issue of NIHL based on a group of 2162 subjects taking part in the NSH,⁹ using statistical modeling techniques. That analysis showed HTLs in noise-exposed subjects that were consistent with other published studies, but noise-free subjects had substantially poorer hearing than the controls that had been used by other studies. Hence, the difference between noise-exposed and noise-free subjects (i.e., NIHL) was smaller than previously reported. None-

theless, the pattern of NIHL with frequency was consistent with previous studies, having a maximum at 4 kHz. This effect was demonstrable in males with occupational noise exposure, but a similar effect could not be shown for females.

Any such modeling study has substantial statistical difficulties to overcome. Particular difficulties arise when dealing with age, which is inevitably collinear with occupational noise exposure, and exerts a powerful influence on HTL, especially in subjects above about 50 years (see Figure 30-3). There are also substantial problems encountered with the distribution of HTL, which varies with age, sex, and occupational group, and is different at each audiometric frequency. At 3 kHz and above the overall distribution of HTLs tends toward the normal distribution, but at lower frequencies the distribution is more like a log-normal distribution. Furthermore, the distributions tend to differ between noise-exposed and noise-free groups. At present, no completely satisfactory statistical method exists to model the data, and the results of any modeling exercise are subject to some uncertainty. In our earlier analysis,⁹ we used the general linear interactive modeling (GLIM) method¹⁶ which allows the building of hierarchical linear models. We partially overcame the problems outlined above by separating male and female data. This allowed the building of satisfactory models to predict HTL from the addition of the main effects of age, occupational group, and occupational noise exposure, with no interactions. Separate models of the same form, but with different parameter values, were required for each audiometric frequency. They explained up to 58% of the variance in HTL, depending on frequency. One unsatisfactory aspect of these models was an inability to show significant dependence of HTL on gunfire noise exposure that we rationalized as due to confounding with age, given that most exposure had been in the World Wars or in military service around that time. Since our earlier publication, we have reexamined the same data with a view to achieving a more satisfactory model. A brief account of the refined model has been given by Lutman,

Davis and Spencer.⁵ The following description is a fuller account of the analysis concentrating on comparisons of the model with data from other studies.

In contrast to the sampling strengths of the NSH, study of NIHL in population samples has weaknesses. In particular, direct measurement of noise levels constituting historical noise exposure is not feasible. As a consequence, noise levels, durations, and exposure patterns must be estimated from reports given by subjects at clinical interview. We have developed a rigorous protocol to make this process as accurate as possible, but any estimates obtained from clinical reports must be subject to a substantial degree of uncertainty. For this reason, we have graded cumulative noise exposure into just four broad bands, consisting of an effectively noise-free band, plus three others that are 10 dB wide. Furthermore, we have been forced to make assumptions about the trade-off between noise level and duration to obtain a composite measure of noise exposure. For this purpose we elected to use NIL, which is based on the equal-energy principle described above. Our data do not permit investigation of the nature of such a trading relationship. Taking together the strengths and weaknesses of our population study approach, the NSH provides evidence complementary to studies based on intensive study of specific noise-exposed groups. It allows an assessment of the extent to which previous intensive studies may have been biased by inappropriate controls, and it provides a set of general-purpose control data that may be used for such purposes. It should be noted that denial of noise exposure by any subjects in the NSH must be equally as valid as in other studies, and hence the noise-free data of the NSH do not have particular weaknesses related to the estimation of noise exposure.

Methods

Subjects were drawn from the NSH, which is a large study covering many aspects of hearing in adults of all ages.¹⁵ The variables of interest here are audiometric thresholds, noise exposure measures, and demographic parameters

of sex, age, and occupational group. The raw data and their classification were described by Lutman and Spencer.⁹ Subjects were selected at random using a two-stage method of stratified sampling that is a standard approach for such surveys and has the advantages over simple random sampling of greater efficiency where the aim is to establish prevalence of specific conditions in the population. Briefly, in the first stage subjects were selected at random from the electoral registers of Cardiff, Glasgow, Nottingham, and Southampton, in three phases between 1980 and 1985, and were sent a postal questionnaire. The second stage was a random sample of the respondents stratified predominantly on age group, self-reported hearing status, and persistent tinnitus, as determined by the postal questionnaire. The second stage systematically oversampled those reporting hearing difficulty, tinnitus, or possession of a hearing aid on the initial postal questionnaire, relative to the general population. Because of the rarity of moderate and severe losses, there was still a preponderance of subjects with normal and mild hearing loss after the second stage of sampling. It was possible to attach a weight to each stratum that reflected the prevalence of that stratum in the population for a particular sex and location. Population attributes were determined using well-established sampling theory¹¹⁻¹⁹; an example of the use of such methods to establish prevalence of hearing impairment exceeding a stated criterion is given by Davis.¹⁵ Further sampling of subjects over a wider geographical area has shown that the prevalences of hearing impairment thus obtained were representative of the general population of the UK.¹⁵ Furthermore, telephone contact with subjects not answering the initial questionnaire, and domiciliary visiting of those not willing to attend the clinics at the second sampling stage, indicated that there were no material response biases, within defined strata based on age and sex. For all analysis that follows, data were weighted to represent population characteristics.

A total of 2708 subjects aged 18-80 years were seen at the clinics between 1980 and 1986. Audiometry was performed in 5 dB

steps using a procedure based on the modified Hughson–Westlake method of manual audiometry recommended by the British Society of Audiology/British Association of Otolaryngologists.²⁰ This is similar to the ascending method described in ISO 8253–1.²¹ Air-conduction thresholds were obtained at 0.25, 0.5, 1, 2, 3, 4, 6, and 8 kHz. Calibration of hearing levels was to the nearest 1 dB relative to British Standard BS 2497,²² technically identical to ISO 389.²³ Middle ear status was characterized by the air–bone gap averaged over the frequencies 0.5, 1, and 2 kHz.

Cumulative noise exposure was assessed retrospectively by structured interview, separately for occupational noise, leisure noise, and gunfire. For the present analysis, occupational and leisure noise immission were classified in four bands corresponding to NILs* of: <97, 97–107, 107–117, and >117 dB(A). This classification differs slightly from that used by Lutman and Spencer⁹ which aggregated the two highest bands. In fact, leisure noise never exceeded an NIL of 107 dB(A), and hence there were effectively only two categories of leisure noise. The effect of any hearing protection worn during noise exposure was accounted for by an appropriate reduction in noise level. Gunfire noise was classified in three bands: <1000, 1000–10 000, and >10 000 rifle rounds, or equivalent for heavier guns. Only unprotected exposure was counted. Further details of these classifications are given by Lutman and Spencer.⁹

To concentrate on NIHL, subjects with other possible ear pathology were excluded, insofar as this was possible. Specifically, ears with any conductive hearing impairment, as defined by an air–bone gap >5 dB, averaged over the frequencies 0.5, 1, and 2 kHz, were excluded. Any subjects with a history of meningitis, sudden or fluctuating hearing loss, or who had received potentially ototoxic drugs,

were excluded. Subjects with an HTL averaged over 0.5, 1, and 2 kHz in the better ear >80 dB were excluded on the grounds that this could not arise primarily from noise exposure, and also to improve the statistical modeling by eliminating extreme cases. We also concentrated on analysis of the better ear to exclude unidentified unilateral pathology, which was unlikely to arise from noise exposure. In addition, subjects over the age of 80 years were excluded owing to the difficulties of achieving representativeness in the elderly population. After all these exclusions there remained 1968 subjects with complete audiograms (944 male, 1024 female). Table 30-1 indicates the numbers of male and female subjects in each occupational group, for each level of noise exposure. As might be expected, the numbers of female subjects in the higher NIL bands were low, as were the numbers of male subjects in nonmanual occupations.

Results

Our general approach aimed to quantify the effects of noise exposure as sensitively as possible while controlling for other effects using a GLIM approach. Exploratory examination of the data showed an unexpected interaction between occupational and gunfire exposure that, although not significant in the analysis reported by Lutman and Spencer⁹ due to small numbers of subjects involved, exerted a material influence on the modeling. The interaction was evident as relatively good hearing in a few subjects with high levels of both occupational and gunfire noise exposure. A factor was included to represent this interaction.

Previously we had modeled the effects of age by separate factors for each 10 year age band without making any assumptions about the underlying shape of the function relating HTL to age. ISO 7029 incorporates a quadratic function of (age–18), where age is in years and a different multiplier is used for each audiometric frequency.¹³ This approach offers the possibility of a more parsimonious description of the data and avoids the effects of quantization in 10 year bands. Exploration of the data

*NIL is numerically equal to the A-weighted level of a noise exposure that lasts for 1 year, or its equivalent based on equal energy. Hence, an NIL of 97 dB(A) could arise from exposure to, for example, 97 dB(A) for 1 year, 94 dB(A) for 2 years, 91 dB(A) for 4 years, 87 dB(A) for 10 years, etc.

Table 30-1 Subjects in Each Occupational Group

NIL Band [dB (A)]	Male		Female		Median NIL [dB (A)]
	Nonmanual	Manual	Nonmanual	Manual	
<97	325	278	473	405	—
97–107	49	147	35	84	101.8
107–117	13	88	4	23	110.5
117+	8	36	0	0	119.8

NIL, noise immission level.

indicated that a quadratic in age rather than (age-18) gave a better fit to the data.

Further exploration and preliminary model fitting also indicated that leisure noise was not significant and that separate models for the two occupational groups could give a more parsimonious fit and less confounding of age, occupational group, and NIL, than a single model. Even when subjects with an air-bone gap >5 dB had been excluded, the model fit was improved by including an air-bone gap term. Hence, based on many trials of different statistical models, our final model took the form shown in Eqn (30-2) for each sex and occupational group.

$$\text{HTL}_i = C_i + A_i \times \text{age}^2 + B_i \times \text{abg} + N_{ij} + G_{ik} + NG_{ijk} + e_{ijkl} \quad \text{Eqn (30-2)}$$

where HTL_i is the HTL in the better ear at the frequency given by the index i ($i = 0, 1, \dots, 7$); C_i , A_i , and B_i are constants for each frequency i ; age is the subject age in years; abg is the maximum air-bone gap over the frequencies 0.5, 1, and 2 kHz; N_{ij} is a factor corresponding to the frequency i and the occupational noise exposure band j ($j = 0, 1, 2$, or 3); G_{ik} is a factor corresponding to the frequency i and the gunfire noise exposure band k ($k = 0, 1$, or 2); NG_{ijk} is a factor corresponding to the frequency i and the conjunction of values of j and k of 2 or greater; and e_{ijkl} is an error term assumed to be normally distributed with zero mean.

We tested other assumptions to stabilize the variance including using a logarithmic transformation. No material difference arose using

the different assumptions. Hence, the exact nature of the distributional assumptions is not critical and our chosen model would appear to be robust, if somewhat underpowered.

The set of models can be summarized for each sex and occupational group by regression constants C , A , and B , and the factors N , G , and NG , having 4, 3, and 2 values, respectively. The constant C effectively incorporates the baseline values of the other factors (having zero index) leaving redundant zeros for the lowest levels of N , G , and NG . There were only three female subjects with any material gunfire noise exposure. Hence, for simplicity, the G and NG factors were excluded from the female model. Also, no females had high levels of occupational noise and therefore factor N had only three levels (N_0 , N_1 , and N_2). Table 30-2 gives estimates for males of constants C , A , and B , and for factors N , G , and NG for each audiometric frequency. Separate sets of estimates are given for manual and nonmanual occupations, with estimates of the variances accounted for by each model. Table 30-3 gives similar data for females. Tables 30-2 and 30-3 are identical to those we have published previously.⁵

Examination of Table 30-2 shows significant effects of the highest noise exposure band (N_3 : NIL > 117) at all frequencies in both occupational groups, with a maximum of 29.9 dB in the nonmanual group at 3 kHz and 20.0 dB in the manual group at 4 kHz. Lesser effects are evident at the immediately lower level (N_2 : NIL 107–117). There are no significant effects at the lowest noise-exposed level (N_1 : NIL 97–

Table 30-2 Model Parameter Estimates for Males and Percentage of Variance Explained

	Frequency (kHz)							
	0.25	0.5	1	2	3	4	6	8
<i>Nonmanual</i>								
C	6.5	1.7	-1.2	-1.7	-2.0	-0.1	8.5	0.1
A × 1000	1.6	2.2	3.0	4.5	6.6	7.9	8.9	11.1
B	0.2	0.5	0.4	0.2	0.0	-0.2	0.4	0.4
N ₁	1.1	1.9	0.6	1.7	2.2	1.7	1.1	0.7
N ₂	0.0	0.0	1.6	1.3	10.4	17.5	14.9	7.8
N ₃	7.7	9.5	10.2	19.6	29.9	24.1	24.7	16.1
G ₁	-5.1	-4.9	-2.9	-2.0	-0.6	-2.8	0.3	3.9
G ₂	-1.0	1.3	2.9	-0.5	-0.6	-4.4	-3.6	1.2
NG	13.3	12.6	16.5	0.6	-8.9	-3.1	1.7	9.1
Var. (%)	15.2	24.9	41.2	45.1	55.0	55.5	55.0	61.7
<i>Manual</i>								
C	6.5	2.6	0.9	-0.3	1.8	2.8	9.5	1.7
A × 1000	2.6	2.7	3.0	5.3	7.9	9.6	11.0	12.5
B	0.3	0.3	0.3	0.5	0.8	0.7	0.8	0.5
N ₁	0.2	0.6	1.1	1.0	1.6	1.6	0.2	-1.4
N ₂	2.9	4.3	3.9	5.4	8.7	10.1	5.0	2.4
N ₃	2.7	4.2	7.4	12.2	16.4	20.0	13.3	10.0
G ₁	-2.0	-1.5	-2.6	-0.8	-0.6	3.3	1.9	3.8
G ₂	0.8	-2.4	-3.3	-3.0	-0.5	5.7	-0.2	2.6
NG	-0.8	-1.3	-4.0	-7.3	-13.9	-21.7	-10.6	-12.8
Var. (%)	23.5	23.5	28.2	36.9	46.9	55.0	54.1	61.6

Numbers in italics denote values not significantly different from zero ($p > 0.05$).

107) at any frequency. This latter level might be accrued by working for 50 years at daily levels between 80 and 90 dB(A), or correspondingly shorter durations at higher levels. Note that this is a mean effect across all exposed subjects and does not deny that some more susceptible individuals will sustain greater damage. Gunfire noise did not demonstrate any meaningful pattern of significant effects. Note that apparently sizeable effects in the table may not be statistically significant due to restricted subject numbers for a particular level of the parameter.

Limited support for the contention that male and female subjects are similarly affected by noise is given by comparison of Tables 30-2 and 30-3. However, shortage of

noise-exposed females, especially in the non-manual group (see Table 30-1), prevents us comparing males and females in more detail in the present analysis.

Taken at face value, the effects of noise in males appear to be somewhat smaller than reported elsewhere. Taking as an example a man aged 60 with an NIL of 110 dB(A), arising from daily exposure to noise at 94 dB(A) for 40 years, the following predictions of NIHL at 4 kHz are obtained from the three methods referred to earlier: NPL method, 25.7 dB; HSE method, 16.0; ISO method and Database A, 18.0 dB. By comparison, Table 30-2 (factor N₂) gives a value of 10.1 (manual, based on 88 subjects) for an NIL between 107 and 117 dB(A). The greatest discrepancy is between

Table 30-3 Model Parameter Estimates for Females and Percentages of Variance Explained

	Frequency (kHz)							
	0.25	0.5	1	2	3	4	6	8
<i>Nonmanual</i>								
<i>C</i>	4.8	0.8	-0.1	-0.9	-0.5	-0.5	7.2	0.2
<i>A × 1000</i>	2.7	3.0	3.3	4.4	5.4	6.4	8.1	9.7
<i>B</i>	0.2	0.5	0.6	0.7	0.5	0.7	0.7	0.7
<i>N₁</i>	1.6	1.4	0.4	-0.9	0.5	-0.8	-0.5	-1.8
<i>N₂</i>	-0.4	-1.4	3.7	6.6	2.4	-4.0	5.5	-2.1
Var. (%)	25.9	28.8	30.9	39.9	44.2	49.4	48.1	52.7
<i>Manual</i>								
<i>C</i>	7.0	3.4	2.6	0.5	1.2	1.4	9.7	0.6
<i>A × 1000</i>	2.5	3.0	2.8	4.2	5.1	6.2	7.9	10.0
<i>B</i>	0.2	0.5	0.1	0.1	-0.1	0.0	0.3	0.4
<i>N₁</i>	-0.8	-0.6	0.0	0.7	1.3	1.6	0.3	-2.8
<i>N₂</i>	2.4	4.0	4.4	5.0	4.0	5.5	3.7	5.6
Var. (%)	18.0	22.2	20.4	33.6	36.1	43.2	45.7	55.0

Numbers in italics denote values not significantly different from zero ($p > 0.05$).

the present model and the NPL method. However, the source of the discrepancy lies primarily in the noise-free data. The direct predictions of HTL for noise-exposed groups are 47.5 and 44.9 dB for the present model (manual occupation) and NPL method respectively, in good agreement. Reference to Figure 30-3 demonstrates that the NPL predictions for HTLs in noise-free groups are the lowest of the five data sets in the figure. This isolated comparison illustrates how estimates of NIHL are strongly influenced by the control data used and emphasizes the importance of careful selection of control data.

Figure 30-4 provides a further comparison between the HTLs predicted from the present model and those predicted from the NPL, HSE, and ISO methods. Each of the three lines in the figure is the predicted median audiogram for a man aged 60 years exposed to noise at a level of 94 dB(A) for 40 years, starting at the age of 20 years [NIL = 110 dB(A)]. The plotted symbols are for the NSH model for a male with a manual occupation and NIL band

107–117 dB(A), calculated[†] using the values given in Table 30-2 which estimate the mean HTL rather than the median. The median NIL for this band is close to 110 dB(A) (see Table 30-1). The NSH model predicts HTLs that lie within the range of the three plotted curves at all frequencies where all three curves provide data (1–4 kHz). The NSH model is particularly close to the NPL curve.

Figure 30-5 compares the NSH model with the same three methods for a noise level of 104 dB(A) for 40 years [NIL = 120 dB(A)]. The plotted symbols are from the NSH model for the NIL band 117+ dB(A), for which the median NIL is close to 120 dB(A) (see Table 30-1). Otherwise parameters are the same as Figure 30-4. In this comparison, the NSH model pre-

[†]Points in Figure 30-4 are calculated using appropriate values of *C*, *A*, and *N₂*. *B*, *G*, and *NG* are set to zero. Hence, at each frequency, HTL is equal to $C + A \times \text{age}^2 + N_2$. For example, at 4 kHz, HTL equals $2.8 + 0.0096 \times 60 \times 60 + 10.1 = 47.46$ for a male with a manual occupation, aged 60 years.

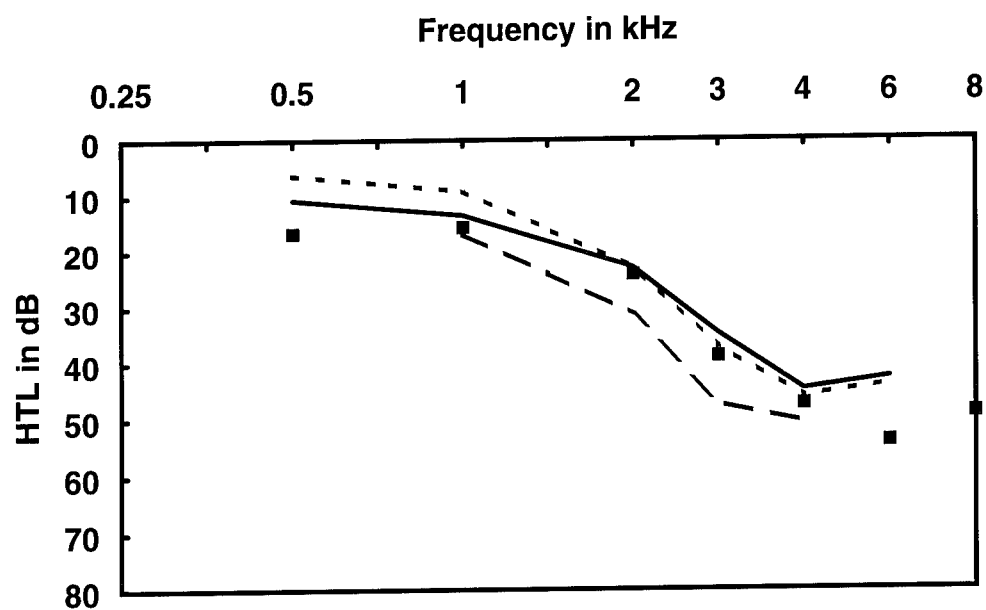


Figure 30-4 Comparison of median hearing threshold levels (HTL) according to three formulas (lines) for males aged 65 years exposed to noise for 40 years at 94 dB(A) [NIL = 110 dB(A)], starting at the age of 20 years, with present statistical model (symbols). Statistical model values are for males aged 60 years in manual occupations exposed to noise immersion levels in the range 107–117 dB(A). (—) NPL Tables.¹⁰ (---) ISO 1999² with Database A.¹³ (- - -) HSE.⁸

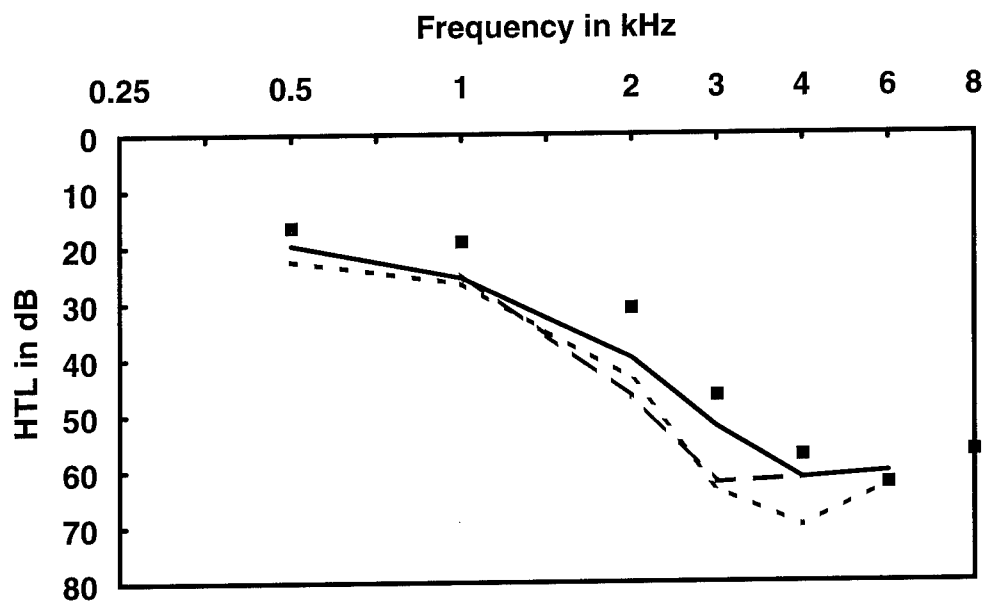


Figure 30-5 Comparison of median hearing threshold levels (HTL) according to three formulas (lines) for males aged 65 years exposed to noise for 40 years at 104 dB(A) [NIL = 120 dB(A)], starting at the age of 20 years, with present statistical model (symbols). Statistical model values are for males aged 60 years in manual occupations exposed to noise immersion levels in the range 117+ dB(A). (—) NPL Tables.¹⁰ (---) ISO 1999² with Database A.¹³ (- - -) HSE.⁸

dicts HTLs that are better than given by the other three methods, especially at frequencies of 1 and 2 kHz where the smallest discrepancies are 6 and 8 dB, respectively. Again, the closest correspondence is with the NPL Tables, the difference amounting to 4 dB at 4 kHz. It should be noted that the NSH only included 43 subjects in this noise exposure band, of whom 34 were classified as having manual occupations. Hence, the estimates given by the NSH model for this band should be treated as surrounded by a substantial range of uncertainty. It should also be recalled that the NSH model relates to the better hearing ear, rather than the average of the left and right ears. Hence, the NSH estimates are expected to be perhaps 3–5 dB lower (better hearing) than given by the other methods. This is consistent with the comparisons shown in Figures 30-4 and 30-5.

Discussion of NSH

The above statistical model was defined with the primary objective of fitting the NSH data, but constrained by certain other criteria. The quadratic function used to represent age originated from other studies that have indicated its general validity. We have resisted the temptation to transform the HTL scale using different functions for each audiometric frequency. That approach could have been used to normalize the error term in the model, but would have entailed some disadvantages. For example, when comparing predicted HTLs at different frequencies, it would be necessary to perform a reverse transformation, making appropriate adjustments to the confidence intervals. Simple use of parametric statistics to compare predicted values would not be appropriate. For these reasons, modeling of the untransformed HTLs was preferred. This choice was borne out by exploratory analysis that indicated a similar outcome of the modeling exercise, independent of the form of the error term used in the model, as described above.

With the above constraints, the model provided a good fit to the data at frequencies above 1 kHz. The percentage of variance in

HTLs explained by the model increases with increasing audiometric frequency. At the frequency of 4 kHz, which is crucial to modeling NIHL, the percentage of variance explained is approximately 55% for males in either non-manual or manual occupations. Both the better fit for males than females, and the increasingly good fit with increasing audiometric frequency, can be accounted for by the greater dependence of HTL on age in males and at higher frequencies, leading to a greater explained variance. This is seen in Tables 30-2 and 30-3 as increased magnitude in the age term in males and at higher frequencies.

The lack of any measured noise levels in the NSH data set does not appear to have led to major discrepancies when comparing with other data sets, at least at the mean/median. The general agreement between the NSH predictions and the NPL, HSE, and ISO formulas, illustrated in Figures 30-4 and 30-5, is generally good. It is certainly as good as comparisons between those formulas. There is a suggestion that the NSH model underestimates HTLs in subjects exposed to very high levels of noise for long periods [$NIL = 120 \text{ dB(A)}$; see Figure 30-5]. This discrepancy may arise from exaggeration of reported noise exposure in some subjects, but the effect is not large. Taking the comparisons of the NSH model and other data for noise-exposed subjects together, there is no suggestion that the lack of noise level measurements has led to a substantial dilution of the noise-exposed group by relatively noise-free individuals. The alternative possibility is that the noise-free group is contaminated by noise-exposed individuals. This could lead to poorer HTLs in the apparently noise-free group than in other studies of noise-free subjects. However, the noise-free subjects of the NSH agree well with other studies at 4 kHz, as shown in Figure 30-3. At other frequencies there is also generally good agreement. Furthermore, we have applied progressively stricter criteria to define the noise-free group in the NSH and observed no improvement in HTL. Therefore, the noise-free group does not appear to be materially contaminated by noise-exposed individuals. Although exaggeration by some subjects can-

not be ruled out, it does not seem to be a common occurrence.

General Discussion and Conclusions

One of the main purposes of this analysis has been to review the steps that are involved in human studies of NIHL and the attendant assumptions that are made. Given the impossibility of discovering what would have been the HTLs of noise-exposed people had they not been exposed to noise, such studies must resort to between-group comparisons. The present analysis has demonstrated that the HTLs of groups of subjects with apparently comparable noise exposures differ widely between studies, even when compared at the median. Likewise, studies of noise-free subjects are far from consistent. Hence, if NIHL is defined as the difference between noise-

exposed and noise-free groups, the numerical values of NIHL obtained will depend heavily on which data sets are selected to represent the two groups. Examination of Figures 30-1 and 30-3 indicates that the NIHL at 4 kHz arising out of a 30 year daily exposure to 100 dB(A) for a man aged 50 years could be anywhere between approximately 10 and 50 dB at the median, depending on the choice of data sets. More extreme examples could probably be found.

Another way to illustrate the extent of overlap between noise-exposed and noise-free groups is to examine the distributions of effects in terms of data from individual subjects. Figure 30-6 plots HTLs at 4 kHz as a function of age for three NIL bands from the NSH. Subjects with an indication of conductive hearing loss, sudden hearing loss, or fluctuating hearing have been excluded. The figure is

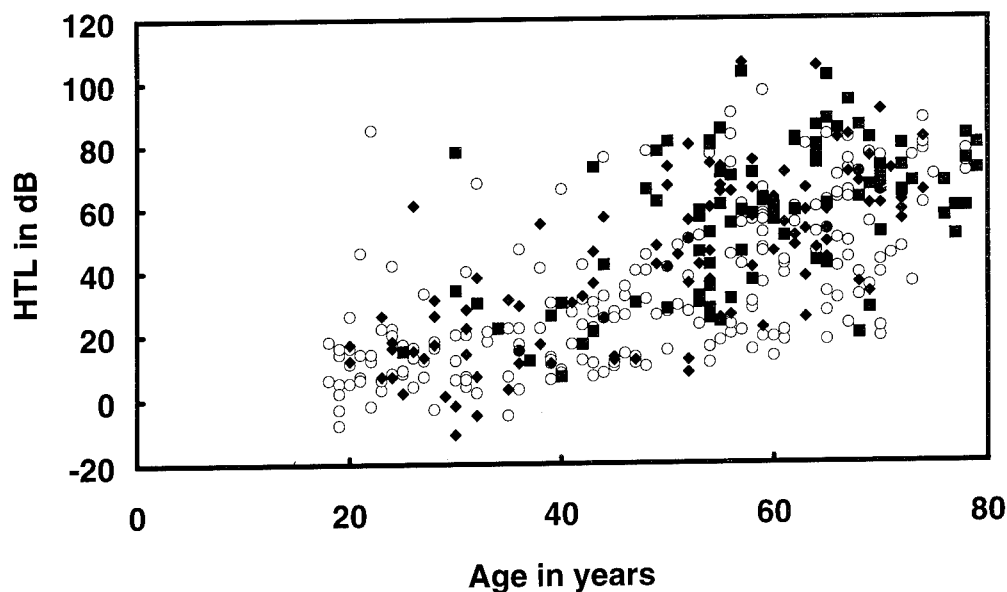


Figure 30-6 Scattergram relating hearing threshold levels (HTL) at 4 kHz in the left ears of males in manual occupations in the NSH to age, with noise immission level (NIL) as the parameter. Subjects over the age of 80 years, with sudden or fluctuating hearing loss, or with air-bone gap averaged over 0.5, 1, and 2 kHz of 15 dB or more are excluded. (○) NIL < 97 dB(A); (◆) NIL 97–107 dB(A); (■) NIL 107+ dB(A).

restricted to males in manual occupations. The NIL bands correspond to <97, 97–107 and >107 dB(A). The overlap between the groups is complete and any distinctions between groups is only meaningful in statistical terms. This illustration emphasizes the fact that any bias in selection of subjects for either the noise-exposed or noise-free groups can have a material influence on group comparisons. Hence, an important consideration when judging the validity of studies of NIHL is the extent to which such bias has been contained. Many studies are silent on this issue.

One possible cause of bias is socioeconomic status. Comparison of the upper and lower halves of Tables 30-2 indicates the effect of occupational group in males. Generally, the coefficients of the age term are greater for males in manual occupations, leading to greater predicted HTLs, even without noise exposure. In our previous analysis, where occupational group was included as a factor,⁹ the effect of manual occupation amounted to an increment in HTL of between 2 and 6 dB in males, dependent on frequency, relative to males with a nonmanual occupation. If studies of NIHL draw their noise-exposed subjects primarily from manual occupations, and their noise-free subjects from nonmanual occupations (e.g. office workers), there will be a confounding of the effects of noise and occupational group. If the difference is attributed entirely to noise, the magnitude of apparent NIHL will exaggerate the true effect of noise. Similar arguments can be applied to other extraneous factors.

The general conclusion from the present analysis is that quantitative statements about the magnitude of NIHL and its dependence on characteristics of noise exposure should be accompanied by an assessment of the robustness of the statements. Few studies have been sufficiently sensitive to extract the fine detail of the underlying relationships between HTL and noise exposure with any degree of confidence. The lack of sensitivity to major differences in assumptions about the way in which the effects of noise and age summation has been illustrated by Bies and Hansen,⁷ who demon-

strated that data in ISO 1999 could be accounted for equally well either by adding decibels directly or on an antilogarithmic basis. (The latter leads to unsustainable conclusions regarding the progression of HTL with increasing age following noise exposure, as pointed out by Macrae²⁴). Theoretical consideration and examination of boundary conditions form an important part of modeling NIHL, in addition to empirical analysis of HTL data. Cross-sectional population studies based on random sampling tend to be insensitive to fine detail of the relationship between HTL and noise exposure, due to lack of power, but provide data that are complementary to more traditional studies concentrating on affected groups. They can allow better control of possible extraneous factors and provide an important cross-check against which traditional studies may be compared. As new statistical methods are developed, it may be possible to increase the power of cross-sectional population studies and to make better allowance for extraneous factors such as age and socioeconomic status.

Acknowledgment

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Chapter 31

Hearing Levels of US Industrial Workers Employed in Low-Noise Environments

William W. Clark and Carl D. Bohl

Currently there exist a number of data bases that describe age-related changes in hearing.¹⁻¹⁰ These data bases fall into two general categories of hearing level data from highly screened (S) and unscreened (U) populations. Data from the S population are useful for determining hearing loss due to age alone (pure presbycusis); they are not generally useful for evaluating occupational noise-induced hearing loss in an experimental population because they underestimate hearing levels of individuals who are unscreened except for occupational exposure to noise.^{3,4}

Numerous studies of hearing levels of unscreened, noninstitutionalized adults have been published and summarized by Robinson.⁵ However, these studies include some individuals who were exposed to occupational noise that may inflate the estimates slightly. Additionally, surveys derived from random samples of the entire population may not be appropriate for comparison to industrial workers because of differing nonoccupational noise exposure histories in the two groups.

The purpose of this investigation was to establish a control population describing age-related changes in hearing sensitivity for American male and female industrial workers who are unscreened except for exposure to occupational noise, that is, an Annex B for American industrial workers. Data were obtained from hearing conservation programs of 22 American and Canadian companies in an ANSI working group 12.12 and provided by

the National Institutes for Occupational Safety and Health² (NIOSH). A second objective was to determine if hearing sensitivity differs in African-American and Caucasian male and female industrial workers.

Methods

Hearing level data for US industrial workers were obtained from a computer tape of industrial audiometric data provided by NIOSH. This tape includes data from 22 companies representing different types of industries and from different areas of the United States and Canada. The original data were collected by ANSI WG12.12 and used to evaluate the effectiveness of industrial hearing conservation programs.¹¹ The data are anonymous with regard to company identifiers and employee identifiers. Within the 22 data sets there are 15,297 employees with at least four audiograms. Variables included in some (but not all) data sets in addition to hearing threshold levels (HTLs) are: age, date of birth, date of employment, gender, race, time-weighted average noise levels, and nonoccupational or medical factors.

The entire data set was read from the original tape onto Macintosh personal computers for further evaluation using commercially available software (Microsoft Excel; Statview II). Because our purpose was to evaluate hearing levels of US industrial workers who were free from significant occupational exposure to noise, we used the following criteria for

Table 31-1 Number of Subjects Included in Comparison of Hearing Levels of Nonindustrial Noise-Exposed Workers and ISO 1999 Annex B

Age (Years)	Frequency (Hz)					
	500	1000	2000	3000	4000	6000
<i>1st test</i>						
Males						
20	395	395	395	395	395	395
30	328	328	328	328	328	328
40	185	185	185	185	185	185
50	138	138	138	138	138	138
60	55	55	55	55	55	55
Total males	1101	1101	1101	1101	1101	1101
Females						
20	168	168	168	168	168	168
30	144	144	144	144	144	144
40	94	94	94	94	94	94
50	72	72	72	72	72	72
60	37	37	37	37	37	37
Total females	515	515	515	515	515	515
Grand total	1616	1616	1616	1616	1616	1616
<i>Annex "B"</i>						
Males						
20	7135	7139	7139	7139	7139	7139
30	10 280	10 281	10 281	10 281	10 281	10 281
40	11 369	11 371	11 372	11 372	11 372	11 370
50	10 031	10 034	10 032	10 034	10 034	10 032
60	7513	7517	7517	7517	7517	7517
Total males	46 328	46 342	46 341	46 343	46 343	46 339
Females						
20	8423	8429	8429	8427	8426	8428
30	11 286	11 290	11 289	11 289	11 290	11 290
40	12 323	12 325	12 325	12 324	12 324	12 325
50	10 534	10 541	10 541	10 542	10 542	10 541
60	8119	8120	8120	8120	8119	8119
Total females	50 685	50 705	50 704	50 702	50 701	50 703
Grand total	97 013	97 047	97 045	97 045	97 044	97 042

selecting individuals to be included in the analysis:

1. US company (two data sets were from Canada);
2. individuals with time-weighted average noise levels at or below 85 dBA (10 data sets included);
3. individuals with at least four audiograms;
4. individuals with the first audiometric test within 24 months of employment.

The number of subjects evaluated after application of the screening criteria listed above are shown in the top panel of Table 31-1.

The control sample of representative HTLs of US adults was obtained from Annex B of ISO Standard 1999. Annex B is an example of a data set representative of US adults not exposed to occupational noise. It is derived from the 1960–1962 US Public Health Survey⁹ of hearing levels of US adults by age and sex. The bottom panel of Table 31-1 lists the number of subjects examined in the US Public Health Survey of 1960–1962.

For the comparisons of hearing level by age, gender, and race, application of the four exclusion criteria listed above resulted in a very small sample of black workers. Therefore, for

Table 31-2 Number of Subjects Included in Comparison of Hearing Levels of Black and White Workers

Age (Years)	Frequency (Hz)						
	500	1000	2000	3000	4000	6000	8000
<i>Blacks</i>							
Males							
20	531	531	531	528	531	529	46
30	399	399	399	395	399	395	60
40	132	131	131	132	132	131	12
50	67	67	67	67	67	67	8
60	30	30	30	30	30	30	7
Total males	1159	1158	1158	1152	1159	1152	133
Females							
20	192	192	192	192	192	192	28
30	161	161	161	160	161	160	42
40	77	77	77	77	77	77	29
50	25	25	25	25	25	25	10
60	3	3	3	3	3	3	3
Total females	458	458	458	457	458	457	112
Grand total	1617	1616	1616	1609	1617	1609	245
<i>Whites</i>							
Males							
20	1322	1322	1322	1303	1322	1303	139
30	1215	1215	1214	1208	1215	1208	96
40	587	587	587	585	587	585	69
50	375	375	374	374	373	374	42
60	79	79	79	79	79	79	9
Total males	3578	3578	3576	3549	3576	3549	355
Females							
20	366	366	366	366	366	366	41
30	336	336	336	336	336	336	47
40	275	275	275	274	275	274	37
50	180	180	180	180	180	180	20
60	70	70	70	70	70	70	8
Total females	1227	1227	1227	1226	1227	1226	153
Grand total	4805	4805	4803	4775	4803	4775	508

comparisons by race, the first audiometric test for all subjects as included in the sample. The number of subjects included in the analysis by race, gender, age, and audiometric test frequency are listed in Table 31-2.

Results

Hearing Levels of US Industrial Workers

Hearing levels (in dB, re: ANSI S3.6, 1969) of 1101 male and 515 female workers as a function of age and audiometric test frequency are listed in Table 31-3. These thresholds repre-

sent the hearing levels of the better ear of workers obtained on the first audiometric test and within the first 2 years of employment; as such they are representative of the hearing ability of US industrial workers when they are hired into a low-noise job. These data are displayed as the 90th, 50th, and 10th percentiles of the distribution of hearing levels for a given age range, gender, and audiometric test frequency. The data are compared to Annex B of ISO 1999 in Figure 31-1 for males, and in Figure 31-2 for females. The data for Annex B have been corrected according to a previous report,¹² and are included in Table 31-4.

Table 31-3 Hearing Levels of New Industrial Workers Employed in Low-Noise Environments

Hearing Threshold Levels (dB, re: ANSI S-3.6)															
		Age (Years)													
		20			30			40			50			60	
		Fractiles													
Frequency (Hz)	0.9	0.5	0.1	0.9	0.5	0.1	0.9	0.5	0.1	0.9	0.5	0.1	0.9	0.5	0.1
Males															
500	5	13	23	4	12	24	5	13	24	3	14	27	7	17	30
1000	1	9	18	1	9	21	3	11	22	4	11	27	7	16	29
2000	0	8	19	0	8	24	2	10	26	2	14	42	6	21	55
3000	0	10	23	1	12	35	5	18	55	10	35	65	15	43	73
4000	1	11	28	2	15	50	7	24	63	15	46	69	23	56	82
6000	3	14	35	5	20	55	10	25	66	20	47	80	28	56	82
Females															
500	2	10	20	5	13	23	5	13	26	5	15	37	10	20	33
1000	-2	6	15	1	8	18	1	9	24	1	10	27	8	19	33
2000	-2	5	15	-2	7	17	1	9	24	0	11	28	6	18	37
3000	-3	6	16	0	8	18	2	12	32	4	15	37	7	22	53
4000	-1	8	19	0	10	20	4	13	33	6	23	45	14	30	43
6000	1	13	26	1	14	28	7	19	42	11	30	51	18	39	56

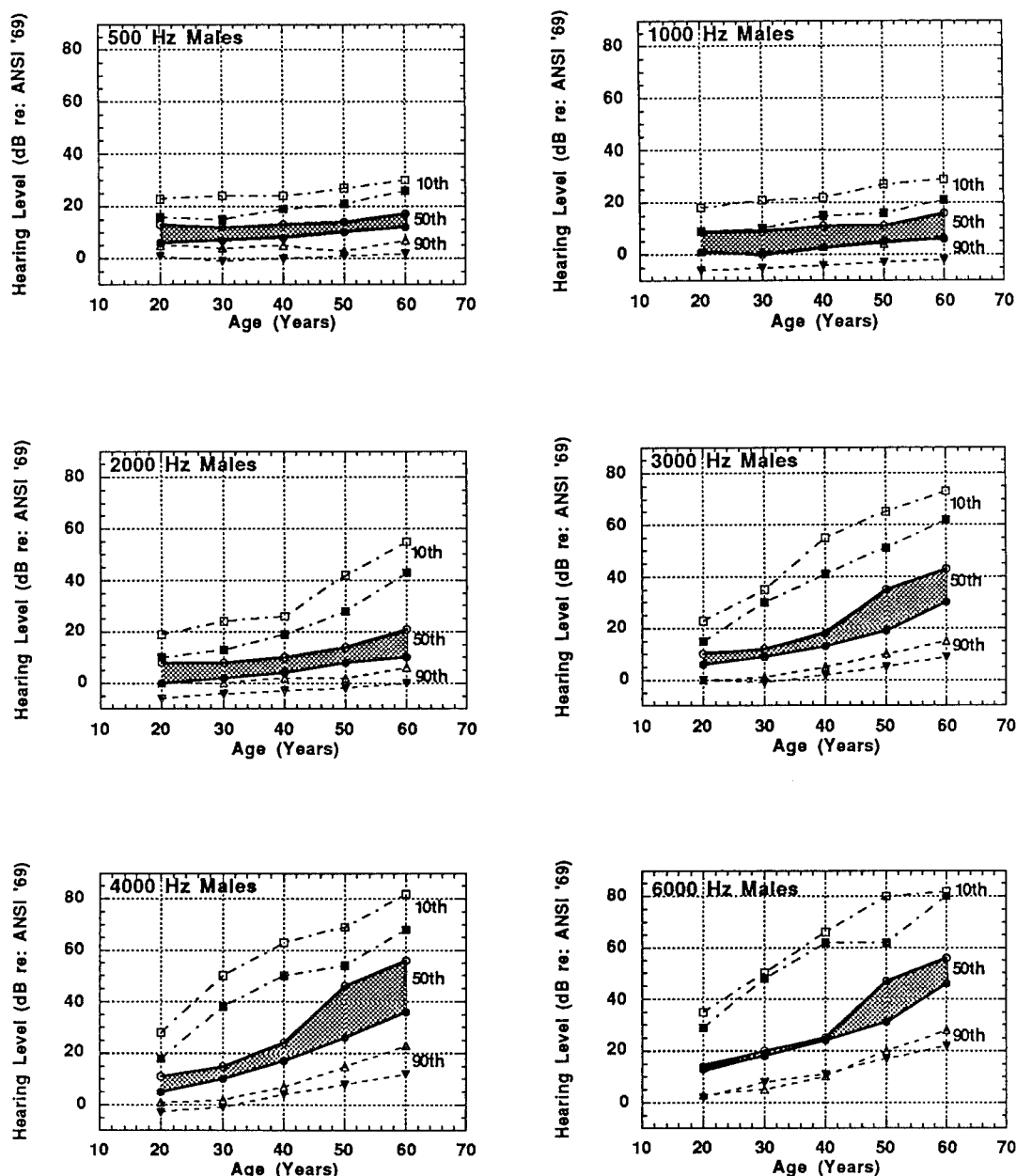


Figure 31-1 Comparison of hearing levels for the better ear of male workers not exposed to industrial noise and Annex B of ISO 1999. Data points plotted on the decile for age (20–60 years) represent age ranges 18–23, 25–34, 35–44, 45–54, and 55–64 years, respectively. (●) Data from Annex B of ISO 1999; (○) data from US industry. Shown are the data for the 10th, 50th, and 90th percentile of the population. Shaded area represents the difference in hearing level for the two groups at the 50th percentile. In this figure, the open symbols (□ ○ △) represent Annex “B” data and filled symbols (■ ● ▲) represent US industry data. Within this breakdown, squares are the 10th percentile, circles are the 50th percentile, and triangles are the 90th percentile.

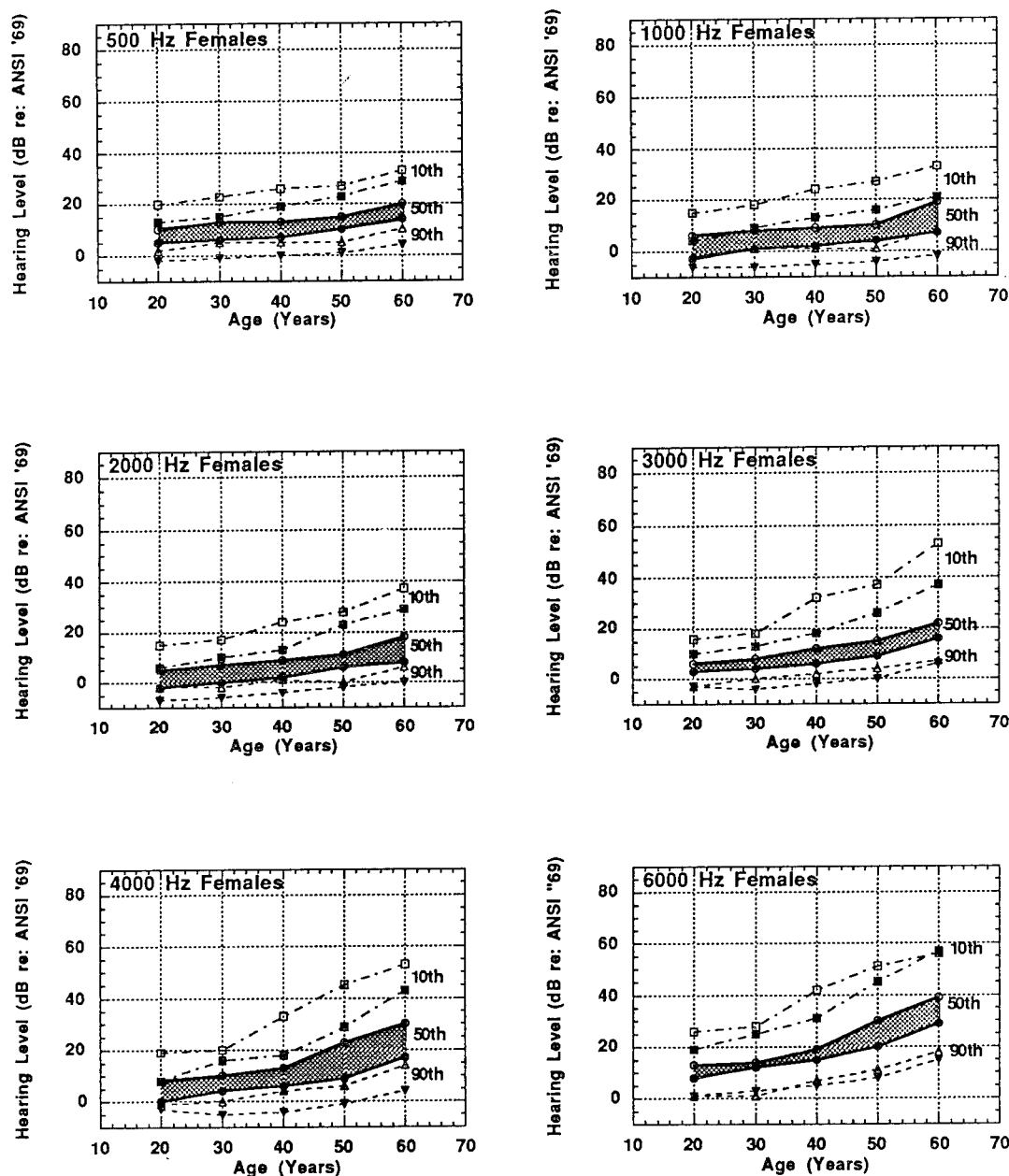


Figure 31-2 Comparison of hearing levels for the better ear of female workers not exposed to industrial noise and Annex B of ISO 1999. Data points plotted on the decile for age (20–60 years) represent age ranges 18–23, 25–34, 35–44, 45–54, and 55–64 years, respectively. (●) Data from Annex B of ISO 1999; (○) data from U.S. industry. Shown are the data for the 10th, 50th, and 90th percentile of the population. Shaded area represents the difference in hearing level for the two groups at the 50th percentile. In this figure, the open symbols (□ ○ △) represent Annex “B” data and filled symbols (■ ● ▲) represent US industry data. Within this breakdown, squares are the 10th percentile, circles are the 50th percentile, and triangles are the 90th percentile.

Table 31-4 Corrected Hearing Level Values of ISO 1999 Annex B

Hearing Threshold Levels (dB, re: ANSI S-3.6)												
Age (Years)												
Fractiles												
Frequency (Hz)	0.9	0.5	0.1	0.9	0.5	0.1	0.9	0.5	0.1	0.9	0.5	0.1
Males												
500	1	6	16	-1	7	15	0	8	19	1	10	21
1000	-6	1	9	-5	0	10	-4	3	15	-3	5	16
2000	-6	0	10	-4	2	13	-3	4	19	-2	8	28
3000	0	6	15	-1	9	30	2	13	41	5	19	51
4000	-3	5	18	-1	10	38	4	17	50	8	26	54
6000	2	13	29	8	18	48	11	24	62	17	31	62
Females												
500	-2	5	13	-1	6	15	0	7	19	1	10	23
1000	-6	-3	4	-6	1	9	-5	2	13	-4	4	16
2000	-7	-2	6	-6	0	10	-4	2	13	-2	6	23
3000	-3	3	10	-4	4	13	-2	6	18	0	9	26
4000	-3	0	8	-5	4	16	-4	6	18	-1	9	26
6000	1	8	19	3	12	25	5	15	31	8	20	45
							</					

Males

Data for newly hired male industrial workers are plotted in Figure 31-1. Each panel represents one test frequency; data for industrial workers are presented as open symbols and data from Annex B are presented as closed symbols. Differences between industrial workers and controls at the 50th percentile are shown by the shaded area. Examination of Figure 31-1 shows that HTLs for newly hired male industrial workers were worse than the controls at all frequencies and for all percentiles. At 500, 1000, and 2000 Hz, the difference was relatively constant as a function of age, and for the 50th percentile it ranged from 7 to 12 dB (shaded area in Figure 31-1). At test frequencies of 3000, 4000, and 6000 Hz, however, differences were smaller for employees in the 20-, 30-, and 40-year-old age groups, but larger for the two older groups. At 4000 Hz, differences in thresholds for the 50th percentile of the male industrial workers compared to those reported in Annex B of the ISO standard reached 20 dB for the 50 and 60 year olds.

Less pronounced differences between industrial workers and Annex B were observed at 3000 and 6000 Hz. However, the trend was similar to that observed at 4000 Hz, namely, a tendency for the differences between the industrial workers and an unscreened sample representative of a non occupationally exposed group of US males to increase with increasing age.

Females

Comparison of hearing levels by percentile for female industrial workers with Annex B of the ISO standard are presented in Figure 31-2. The data are plotted in the same fashion as Figure 31-1. Like the males, female industrial workers tended to have worse hearing than that reported as typical for US females at all test frequencies. The size of the difference was typically 7–8 dB for the 50th percentile, but was as large as 13 dB for 60-year-old females at 4000 Hz. Unlike the data for the males, hearing sensitivity of older females did not differ from controls by as much at 3000, 4000, and 6000 Hz.

Differences in hearing sensitivity between females and males in this sample were about the same as those observed in Annex B of the ISO standard, females tended to have better hearing than males at all frequencies, age groups, and percentiles.

Racial Differences in Hearing: Thresholds of Black and White Male and Female Industrial Workers

Of the 22 data bases included in the ANSI file, 7 included information on racial characteristics of the workers. Because application of the restrictions listed above for the comparison of industrial workers with Annex B of ISO 1999 resulted in a restricted sample of black workers, it was decided to remove two restrictions: first test less than 2 years after employment, US company, and four tests for each individual for this analysis. Therefore, the resultant data set represents HTLs for the first audiometric test of black and white male and female workers. Like the data described in the previous sample, time-weighted average noise exposure levels were ≤ 85 dBA for the current year for these workers.

The data are presented in Tables 31-5 and 31-6 for males and females, respectively. Figure 31-3 is a graphic comparison of hearing levels of black and white male workers as a function of test frequency. Data are presented for the 10th, 50th, and 90th percentile. HTLs for black workers are shown as filled symbols and white workers shown as open symbols. The shaded area represents the difference between groups at the 50th percentile. At test frequencies below 3 kHz, HTLs differed little between groups; at the 50th percentile the differences were 2 dB or less for all frequencies and age groups. However, the threshold functions differed between groups for the 3, 4, 6, and 8 kHz frequencies.

At those frequencies, younger individuals displayed similar hearing ability, regardless of race. However, hearing levels of white males were worse than those for black males for all ages above 30, and the difference increased with increasing age. For example, thresholds at the 50th percentile for 20-year-old black and

Table 31-5 Hearing Levels of Nonindustrial Noise-Exposed Black Workers

Hearing Threshold Level (dB, re: ANSI S-3.6)												
Age (Years)												
Fractiles												
Frequency (Hz)	0.9	0.5	0.1	0.9	0.5	0.1	0.9	0.5	0.1	0.9	0.5	0.1
Males												
500	3	10	19	3	10	20	4	13	21	3	10	25
1000	1	7	15	1	8	18	2	9	23	2	9	20
2000	1	8	18	0	8	19	2	9	23	5	13	30
3000	1	9	18	1	9	24	4	14	36	5	18	53
4000	1	9	20	2	11	27	6	17	47	7	27	60
6000	1	11	23	2	14	29	2	20	46	11	24	53
8000	0	4	18	0	7	26	2	15	39	2	10	45
Females												
500	2	4	18	1	9	19	2	11	19	3	14	18
1000	0	5	13	1	6	15	0	7	16	-2	8	15
2000	0	6	17	-1	6	15	0	6	15	-1	8	17
3000	1	7	14	0	7	15	1	8	18	-1	9	18
4000	1	8	20	1	8	19	2	9	20	2	11	23
6000	1	11	23	1	10	19	2	12	26	-3	17	31
8000	0	7	21	1	9	22	1	10	27	1	15	20

Table 31-6 Hearing Levels of Nonindustrial Noise-Exposed White Workers

Hearing Threshold Level (dB, re: ANSI S-3.6)												
Age (Years)												
Fractiles												
Frequency (Hz)	0.9	0.5	0.1	0.9	0.5	0.1	0.9	0.5	0.1	0.9	0.5	0.1
Males												
500	5	12	20	5	12	23	5	13	25	5	13	28
1000	1	8	17	2	9	18	2	11	23	4	12	27
2000	0	7	17	1	8	20	2	11	28	3	15	50
3000	1	9	21	2	13	33	6	20	55	10	32	69
4000	2	10	28	4	15	50	8	27	62	15	45	75
6000	4	14	33	6	20	52	11	29	65	15	42	80
8000	1	10	23	0	12	44	7	21	53	12	45	95
Females												
500	2	8	16	2	9	18	2	11	22	2	10	25
1000	0	5	13	0	6	15	1	8	18	0	8	27
2000	-1	5	13	-1	6	15	0	7	20	0	7	35
3000	0	5	14	1	7	20	1	9	25	1	13	45
4000	0	6	15	1	9	23	2	10	28	3	16	50
6000	1	10	22	3	13	26	4	15	31	1	21	44
8000	1	10	21	0	12	23	4	14	37	10	22	64

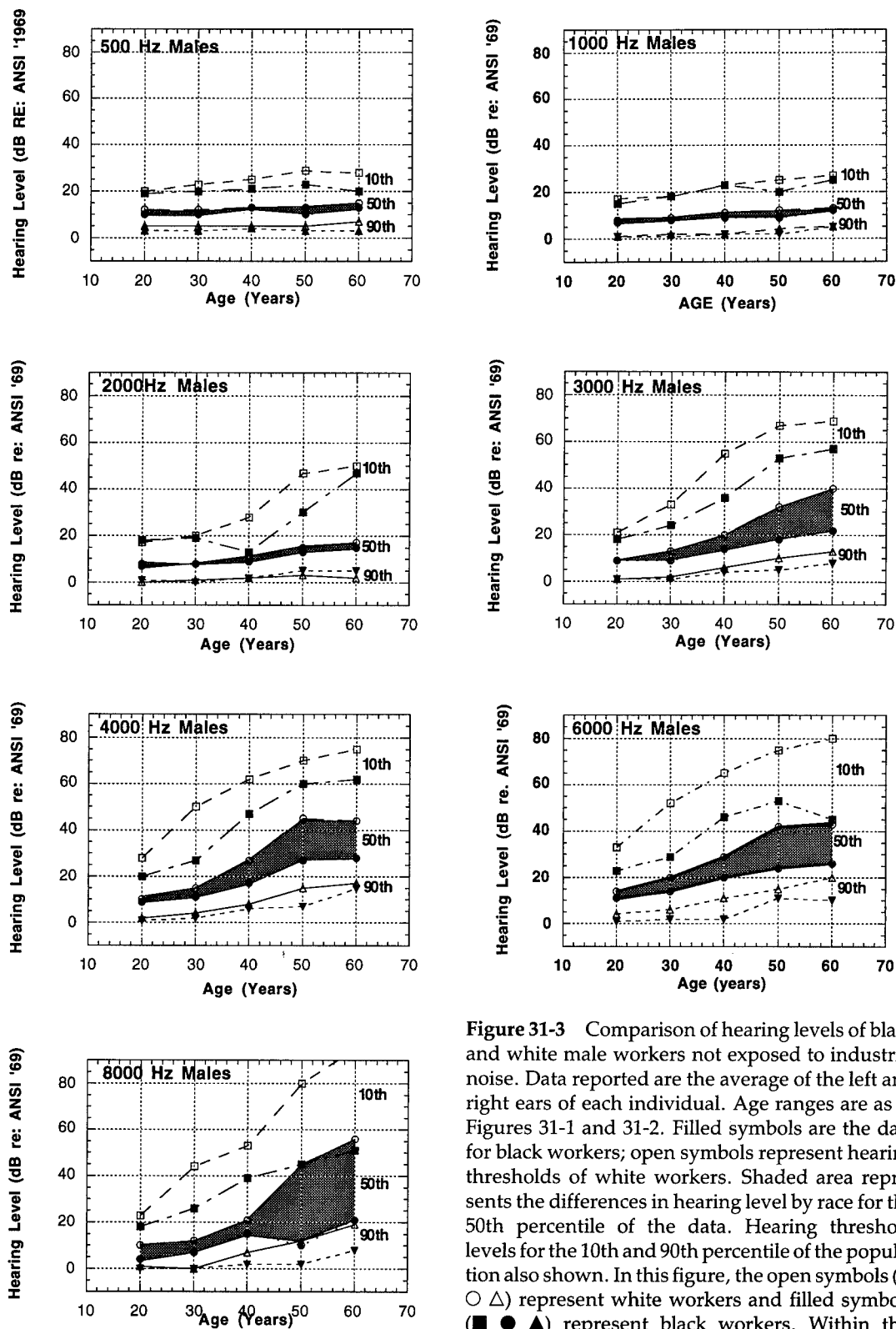


Figure 31-3 Comparison of hearing levels of black and white male workers not exposed to industrial noise. Data reported are the average of the left and right ears of each individual. Age ranges are as in Figures 31-1 and 31-2. Filled symbols are the data for black workers; open symbols represent hearing thresholds of white workers. Shaded area represents the differences in hearing level by race for the 50th percentile of the data. Hearing threshold levels for the 10th and 90th percentile of the population also shown. In this figure, the open symbols (\square \circ \triangle) represent white workers and filled symbols (\blacksquare \bullet \blacktriangle) represent black workers. Within this breakdown, squares are the 10th percentile, circles are the 50th percentile, and triangles are the 90th percentile.

white workers at 4 kHz were within 1 dB of each other (9 and 10 dB HTL, respectively), but White workers had poorer hearing than Black workers by 4 dB at 30 years of age, 10 dB at 40 years of age, and 18 dB at 50 years of age. Data were similar for test frequencies of 3, 6, and 8 kHz.

A similar comparison of hearing levels by age, race, and test frequency for female workers is shown in Figure 31-4. Because the sample size for 60-year-old black females was so small ($n = 8$), they were not included in the comparison. The comparison of hearing levels for females by race for all age groups and test frequencies show little differences between black and white females. Differences at the 50th percentile were typically 0–2 dB, and only at 8 kHz in the 50-year-old group was the difference larger than 5 dB. Unlike the data shown for males in Figure 31-3, data for female workers did not show any significant differences by race.

However, females both black and white have slightly better hearing sensitivity at all test frequencies than males. For example, median HTLs for 50 year olds were: 9 dB HTL for black females; 13 dB HTL for white females; 18 dB HTL for black males; and 32 dB HTL for white males.

Discussion

Limitations of Study

Before discussing the implications of the results, it is important to consider the many weaknesses of a study based upon survey data obtained by voluntary contributions of industrial audiometric data bases. First, the purpose of the project for which the ANSI data base was collected was to compare changes in hearing sensitivity from year to year for individual employees for purposes of evaluating the effectiveness of industrial hearing conservation programs; it was not intended as a source of hearing level data for industry *per se*. Although the transcription of the individual data sets into the NIOSH tape was tested for *accuracy*, no tests were reported on the *quality* of the individual data contrib-

uted by each industry. Therefore, it is reasonable to view the collected hearing level data with some skepticism.

A second, and important limitation was that only the current noise exposure information was collected for each subject; there was no way to ascertain or refute a hypothesis that the workers studied had a prior occupational noise history, even though they currently are not working in noise above 85 dBA. To minimize this uncontrolled variable, it was decided to limit the sample of hearing levels evaluated to the first audiogram obtained within the first 2 years of employment. As such, the audiometric data presented here represent typical hearing levels of new-hire employees who are employed in the current year in low-noise jobs.

It should be recognized that the sample may include individuals with prior occupational noise exposures who are newly hired at midcareer and are currently categorized as low noise exposed. However, prior occupational noise exposure is a part of the history a worker brings to a new job, and therefore should be considered in an evaluation of "typical" hearing levels of industrial employees at the time they are hired.

Finally, although the data came from companies distributed across the United States, a large percentage of the workers included in the data bases came from the South and Southeastern portions of the United States (Figure 31-5 and Table 31-7).

Despite these limitations, the collection of hearing level data contained in the ANSI data base is a valuable resource. It is suggested that the data presented in this chapter should be considered representative of hearing levels of US industrial workers.

Hearing Levels of US Industrial Workers

The data in Figures 31-1 and 31-2 show clearly that men and women working in US industry have hearing levels that are worse than those reported as typical for an unscreened US population. The differences are largest for older males, and at the frequencies usually affected by exposure to noise: 3000, 4000, and 6000 Hz.

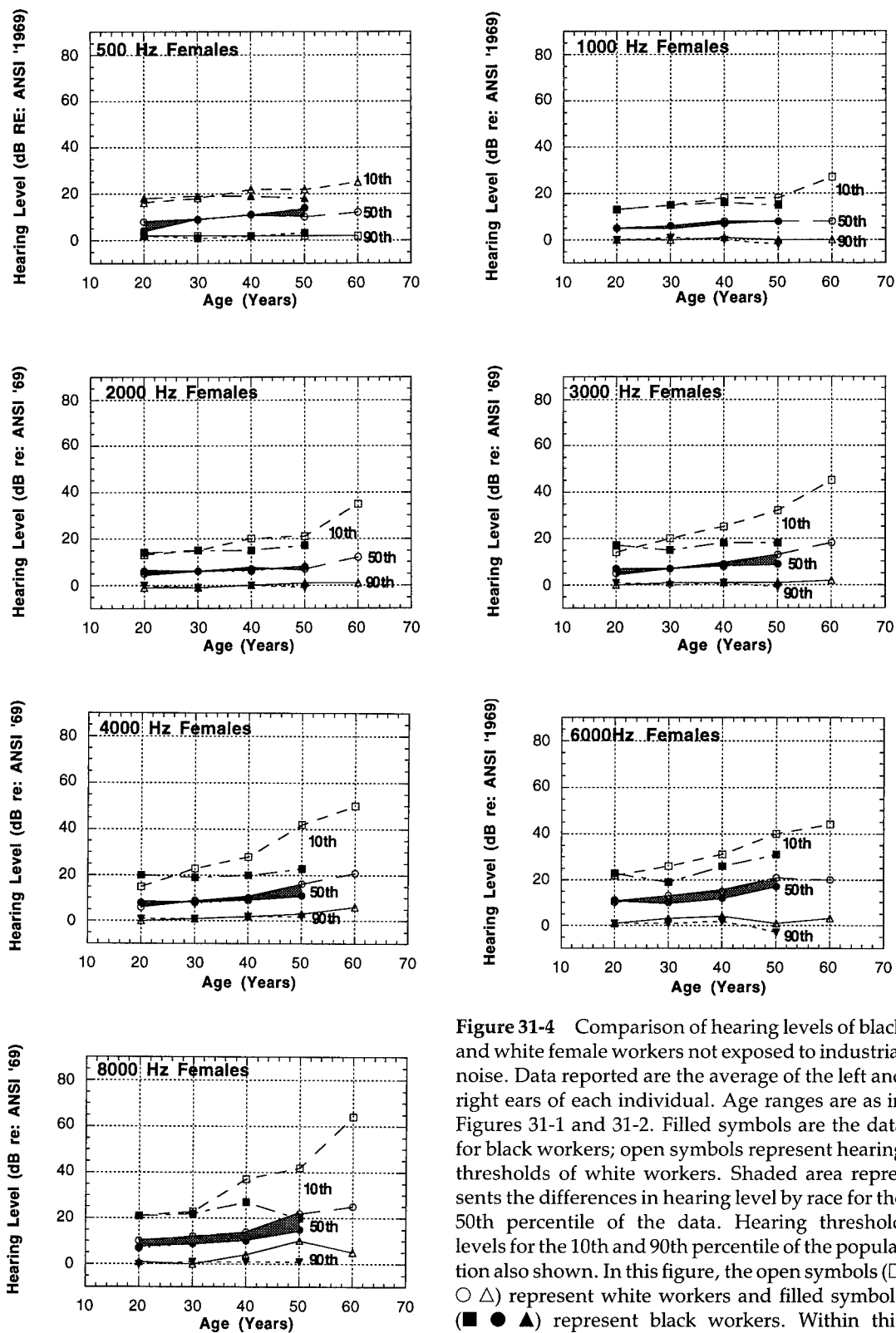


Figure 31-4 Comparison of hearing levels of black and white female workers not exposed to industrial noise. Data reported are the average of the left and right ears of each individual. Age ranges are as in Figures 31-1 and 31-2. Filled symbols are the data for black workers; open symbols represent hearing thresholds of white workers. Shaded area represents the differences in hearing level by race for the 50th percentile of the data. Hearing threshold levels for the 10th and 90th percentile of the population also shown. In this figure, the open symbols (\square \circ \triangle) represent white workers and filled symbols (\blacksquare \bullet \blacktriangle) represent black workers. Within this breakdown, squares are the 10th percentile, circles are the 50th percentile, and triangles are the 90th percentile.

DISTRIBUTION OF NON-NOISE EXPOSED WORKERS

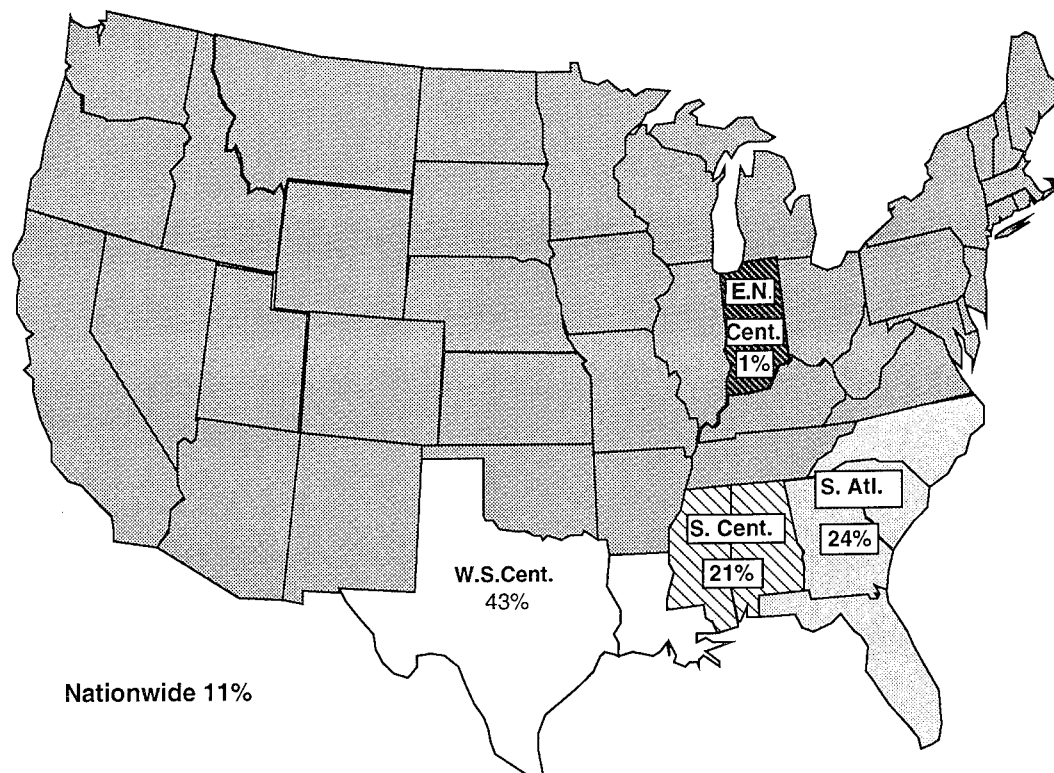


Figure 31-5 Map showing geographic distribution of industrial employees from which comparison of data for race were drawn. Location, type of industry, and percentage racial composition are shown in Table 31-7.

Table 31-7 Racial Makeup, Industry, and Location of Workplace

DB	Black				White			
	M	%M	F	%F	M	%M	F	%F
2	104	8.97	157	34.28	587	16.37	509	41.35
4	601	51.81	91	19.87	1817	50.68	267	21.69
5	41	3.53	33	7.21	24	0.67	36	2.92
6	8	0.69	3	0.66	40	1.12	16	1.30
7	109	9.40	9	1.97	582	16.23	20	1.62
8	297	25.60	165	36.03	532	14.84	383	31.11
19	0	0.00	0	0.00	3	0.08	0	0.00
Grand total	1160	100.00	458	100.00	3585	100.00	1231	100.00

DB, data base; M, male; F, female

Smaller, but important differences occur at all frequencies for males and females.

Some of the observed differences are undoubtedly due to differences in audiometric test procedure. After all, the data included in Annex B were obtained with manual audiometric techniques carried out in a sound-shielded environment. Many of the industrial audiograms, on the other hand, were collected with microprocessor audiometry in less than ideal environments. On the basis of a previous study¹³ it was expected that the hearing levels so obtained would be worse than those reported in Annex B, and they were.

However, the differences due to test environment, audiometric technique, and the fact that these were the first audiometric tests were expected to be on the order of 3–4 dB and to be independent of test frequency, gender, and age. The fact that levels were 7–8 dB worse than the control population suggests that demographic, etiologic, and nonoccupational noise exposure factors were different in these industrial workers, which resulted in poorer hearing.

Because hearing sensitivity of new-hire employees seems to be worse than that observed in a random sample of the US population, any assessment of predicted risk of occupational noise-induced hearing loss for exposed groups of individuals based upon data from ISO 1999, using either Annex A or B as the

control sample, will result in an overprediction for these subjects. In other words, newly hired industrial employees already have the equivalent of 7–8 dB of noise-induced permanent threshold shift (NIPTS) prior to starting work at the current company. It is suggested that the data contained in this chapter provide a more appropriate "data base B" for US workers than the currently recommended Annex B.

Racial Differences in Hearing Sensitivity

Whether there are inherent differences in hearing sensitivity between whites and blacks remains controversial. Based largely upon a sample obtained from a shopping mall and at a trade school in North Carolina, Royster and Thomas¹⁵ suggested that hearing sensitivity of African-Americans, particularly males, is better than that of Caucasians. In fact, Royster has suggested that hearing level data be segregated by race as well as gender, and that separate control data be used for black and white workers.

Kryter¹⁶ thoroughly reviewed all the extant data concerning racial differences in hearing and concluded that there are no demonstrable differences in hearing levels between blacks and whites, either females or males. Kryter argued that the differences observed by Royster and Thomas¹⁵ were undoubtedly due to

Population						Location	Industry
Black	% Black	White	% White	Total	% Total		
261	16.13	1096	22.76	1357	21.09	So. cen.	Syn. fib. mfg.
692	42.77	2084	43.27	2776	43.15	W. so. cen.	Mfg. oilf. mach.
74	4.57	60	1.25	134	2.08	So. Atl.	Meat pack.
11	0.68	56	1.16	67	1.04	E. n. cen.	Mfg. acous. foam
118	7.29	602	12.50	720	11.19	Nationwide	Pet. refin.
462	28.55	915	19.00	1377	21.40	So. Atl.	Text. mill
0	0.00	3	0.06	3	0.05	BC Can.	Al smelt
1618	100.00	4816	100.00	6434	100.00		

different amounts of sociocusis and nosocusis between groups. Kryter's position was also supported by Robinson,⁵ who discounted differences in thresholds observed for "melanoderms" and "leucoderms."

The data reported in this study clearly support the conclusion that there are no inherent differences in hearing sensitivity between black and white workers. The two principle findings displayed in Figures 31-3 and 31-4 are: hearing sensitivity of the 20-year-old groups of blacks and whites are remarkably similar, even though the females of both races tend to have slightly better hearing; and with age, hearing of white males deteriorated more and faster than the other three groups, and the deterioration occurred only at the "noise" frequencies of 3000, 4000, and 6000 Hz.

It is argued that the observed differences in hearing sensitivity are related to the high prevalence of hunting and target shooting among white, male industrial workers, and the very low prevalence in black workers. It is well known that individuals exposed to gunfire may sustain hearing loss associated with those exposures. Reported peak sound pressure levels (PSPL) from rifles and shotguns have ranged from 132 to 139 dBA for .22 caliber rifles,¹⁷ 150–165 dBA reported for 12 gauge shotguns,¹⁹ to 163–170 dBA peak for Mauser pistols.¹⁸ Odess²⁰ has provided a detailed analysis of noise produced by various types of sport rifles and shotguns. He found levels ranging from 163.2 to 172.5 dB PSPL for shotguns, and levels from 143.5 to 170.5 dB PSPL for rifles. Clinical reports concerning hearing loss following exposure to shooting can be found in the literature since the 1800s: Toynbee²¹ noted an association of asymmetrical high-tone hearing loss in patients engaging in the sport of shooting. He also properly identified asymmetrical pattern of exposure caused by shouldering the gun on the right, thus producing a head shadow that protects the right ear.

Numerous studies have attempted to assess the prevalence of hunting or target shooting in the general population. Estimates range from 14% in Scandinavian countries and in the

United Kingdom^{18,19} to 69% in the United States.²² An assessment of shooting history as part of a company hearing conservation program in Canada²³ indicated that 49.5% of the work force responded positively to questions concerning hunting, target shooting, or pistol shooting. These results are generally consistent with other findings from US industry.²⁴

Johnson and Riffle²⁵ evaluated the hearing levels of 68 pairs of workers, matched for sex, age, and exposure level, selected from the Inter-Industry Noise Study. For each pair, one of the individuals had indicated exposure to nonoccupational gunfire during the previous year. Differences in mean hearing level between male subjects exposed to gunfire and those not exposed were clearly apparent and varied between 9 and 16 dB for the frequencies of 3000, 4000, and 6000 Hz. No significant differences were found in thresholds of female shooters; this was attributed to the fact that most females fired guns of small caliber (.22), while most males tended to use several types of guns of larger calibers. Johnson and Riffle concluded that nonoccupational exposure to noise may be a significant problem for men and can be considered equivalent in effect to an occupational exposure of 89 dBA, 8 h/d, for 20 years.

Our interpretation of the data presented is that the differences in hearing level observed between white and black male workers is due to a vastly different exposure history with regard to hunting and target shooting. This hypothesis, however, has proved difficult to test. Attempts to assess membership in the National Rifle Association by race were unsuccessful; similarly, we were unable to find any state that recorded race data on hunting license applications. However, discussions with large-volume distributors of hunting licenses in Missouri confirmed our impression that hunting and target shooting are not popular among black males, but are very popular among rural white males.

Furthermore, because a large portion of the sample came from the Southeastern portion of the United States (Table 31-7), our data on hearing level differences between blacks and

whites may not be representative of the entire US population, as suggested by Kryter.¹⁶ Nevertheless, our findings clearly indicate that there are no inherent differences in hearing sensitivity between black and white Americans.

Conclusion

Hearing levels of US industrial workers were worse than those reported in Annex B of ISO 1999. The difference was larger for males than females, and was as much as 20 dB at 3000, 4000, and 6000 Hz for males over age 40.

Hearing levels of young individuals differed little by gender or race. However, with increasing age, hearing in white males deteriorated more severely at 3000, 4000, and 6000 Hz with age than that observed in the other groups. By age 60 the median HTL at 4000 Hz for white males was 16 dB worse than for black males.

These findings show that US industrial workers have worse hearing than that observed in a random sample from the US population, and that white males lose more hearing at 3000, 4000, and 6000 Hz as they age. The principal factor in causing the additional loss is exposure to noise from hunting and target shooting, an avocation enjoyed by 50–70% of the white male workers, but by few blacks or women.

Acknowledgment

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Chapter 32

Estimation of Occupational Contribution to Hearing Handicap

Robert A. Dobie

In some medical-legal situations, it is necessary to provide estimates of the relative contributions of occupational noise and other causes to an individual's hearing loss. A simple method for using the ISO-1999 models to calculate these estimates will be described and illustrated.

The International Organization for Standardization's publication in 1990 of ISO-1999¹ provided the hearing science community with useful models for the growth of both noise-induced hearing loss (NIHL) and age-related hearing loss (ARHL), as well as a method for combining them to predict hearing threshold levels (HTLs) in noise-exposed populations. Summarizing data from several epidemiological studies, ISO-1999 models noise-induced permanent threshold shift (NIPTS) as a set of logarithmic functions of exposure duration, one for each combination of percentile, exposure level (up to 100 dBA) and audiometric frequency. For a given percentile, exposure duration, and frequency, NIPTS grows as the square of the difference between time-weighted exposure level and a threshold level. For example, at 2 kHz the threshold level is 80 dBA, below which no NIPTS is predicted. Age-related permanent threshold shift (ARPTS) is presented in four data sets: Databases A and B, each of which treat men and women separately. Database A, representing populations highly screened to exclude NIHL and other otologic disease, explicitly models ARPTS as a set of quadratic functions of the difference between actual age and a threshold age of 18 years (one function

for each audiometric frequency and percentile). Database B is derived from the (unscreened) 1960-1962 US Public Health Service survey; although only presented in tabular form, these data are also well-fit by quadratic functions.²

ISO-1999 was designed for *prospective* use in predicting the distributions of HTLs to be expected in a population of specified age, gender, exposure level, and exposure duration. Given these input variables and the audiometric frequency of interest, the model produces a different HTL for each percentile point from the 5th, through the median, to the 95th percentile. The method of combination is simple: find the ARPTS and NIPTS values for the desired percentile, add them, then subtract a "compression factor":

$$\text{HTL} = \text{ARPTS} + \text{NIPTS} - \frac{(\text{ARPTS})(\text{NIPTS})}{120}.$$

Obviously, it is impossible to predict an individual's exact thresholds after a specified exposure. ISO-1999 recognizes this in stating: "This International Standard . . . shall not be used to predict or assess the hearing impairment or hearing handicap of individual persons."

ISO-1999 can, however, support useful probabilistic conclusions about thresholds in individuals. Prospectively, we can predict that an individual's HTLs are likely to fall within a particular range with a specified level of confidence, or that it is highly unlikely that those HTLs will exceed a particular amount.

Given an individual's audiogram, we can also use ISO-1999 for *retrospective* analysis. For example, consider a 60-year-old man whose threshold at 500 Hz is 90 dB hearing level (HL) after working 40 years in a 90 dBA workplace. ISO-1999 formulas predict that 95% of men similarly exposed will have 500 Hz thresholds better than 30 dB HL, suggesting that we must look elsewhere for the cause of this man's hearing loss. ISO-1999 also recognizes this use: "... in doubtful individual cases, the data in this International Standard might provide an additional means for estimating the most probable causes in audiological diagnosis."

Previous publications²⁻⁴ have discussed methods of retrospective inference using the ISO-1999 models. These methods of allocation permit estimates of the relative contributions of NIHL and ARHL in individual medical-legal cases; for example, we might estimate that a worker's loss was 62% due to aging and 38% due to occupational noise. After following the mathematical pathways dictated by different assumptions about ARPTS and NIPTS, a very simple method emerged as probably the best in most circumstances:

1. decide which frequency (or combination of frequencies) is of interest, for prediction of hearing handicap (or "hearing disability," according to WHO terminology⁵);
2. determine whether the individual's thresholds for these frequencies exceed the range (5th–95th percentiles) for which ISO-1999 purports to be valid (steps 3–5 depend on this condition);
3. find the median expected ARPTS for the individual's age and gender (for the frequency or frequencies of interest);
4. find the median expected NIPTS for the individual's exposure level and duration; and
5. allocate the individual's loss to ARHL and NIHL in the same proportions as median ARPTS to median NIPTS.

The purpose of this chapter is to present the reasoning behind this surprising result (sur-

prising because it yields a proportional allocation that is independent of the individual's threshold) in a *graphical* format, rather than the mathematical arguments offered in previous works.

Combining (and Separating) ARPTS and NIPTS

Measured Distributions

Epidemiological studies of NIHL typically have compared pure-tone thresholds for noise-exposed workers (representing the effects of noise plus aging) to thresholds for non-noise-exposed control subjects of matched age and gender (representing age alone). Figure 32-1 shows the results of a hypothetical field study in which 1 kHz thresholds were measured for two groups of 65-year-old men: one group had worked for 40 years at 100 dBA, while the other had had quiet occupations (these curves were drawn using ISO-1999 model parameters). The actual distributions would be skewed; ISO-1999 models this by providing different standard deviations above and below the median values. We have used the upper-half standard deviations for both halves of the distributions for two reasons:

65 y.o. males, 40 yrs. @ 100 dBA

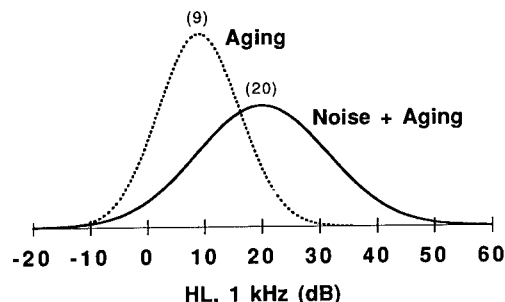


Figure 32-1 Distributions of 1 kHz thresholds, measured in dB hearing level (HL) for two hypothetical populations: noise-exposed and non-noise-exposed (ISO-1999). See text for explanation.

first, it makes our figures easier to interpret; and second, we will confine our discussion to the upper half of the distributions (the vast majority of claimants for compensation have thresholds above median levels). Because of the skew, the points where these distributions peak are medians, not means.

Median threshold for the non-noise-exposed group is 9 dB HL, compared to 20 dB for the noise-exposed group. In addition, the latter distribution is wider; noise exposure increases both mean threshold and variability.

Derived Distributions

How do we represent the effects of noise exposure on thresholds? Imagine that every individual in the "aging" distribution in Figure 32-1 received, in addition to his age-related threshold shift, an additional noise-induced shift, such that the new distribution of thresholds is as shown in the "noise and aging" curve. Obviously, each individual does not receive exactly the same fixed NIPTS; if that were true, the entire curve would simply shift to the right by that amount, without a change in the width of the distribution. Therefore, we must assume that some receive larger NIPTS increments than others, that is, some are more susceptible to NIHL than others.

If NIPTS itself has a mean and standard deviation, we must address the issue of whether susceptibilities to ARPTS and NIPTS are correlated. The simplest assumption, for purposes of calculation, is that they are perfectly and positively correlated. This would mean that the individual unlucky enough to have the largest ARPTS would also be the most susceptible to noise and receive the largest NIPTS. The median individual for ARPTS would receive a median ARPTS, and so on. For each individual, his ARPTS percentile would be identical to his NIPTS percentile. Implausible as this sounds, it makes calculation of an NIPTS distribution very easy, and in fact this is the way that reported NIPTS distributions have usually been derived.

Median NIPTS in ISO-1999 is simply the difference between the median of the noise

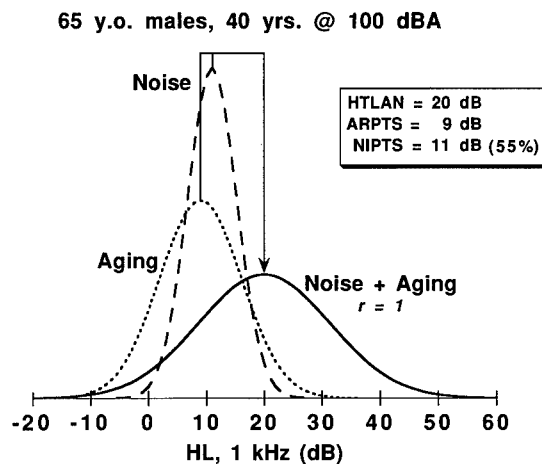


Figure 32-2 A NIPTS distribution has been calculated, assuming perfect correlation between ARPTS and NIPTS, by subtracting corresponding percentile points on the two previous distributions. The fine vertical lines indicate the most likely combination of ARPTS (9 dB) and NIPTS (11 dB) for an individual with a 20 dB HL threshold. HTLAN is ISO-1999's abbreviation for "hearing threshold level due to aging and noise."

and aging distribution and the median of the aging distribution (in the example of Figure 32-1, median NIPTS = 20 - 9 = 11 dB). The 90th percentile NIPTS is the difference between the 90th percentile values of the two measured distributions, and so on. Using this "perfect correlation" ($r = 1$) model, the standard deviation of the NIPTS distribution will be equal to the difference between the two measured standard deviations. Figure 32-2 shows such an NIPTS curve superimposed on the aging and noise and aging curves.

Combining ARPTS and NIPTS

Suppose we are now interested in predicting the distribution of thresholds in a population of 65-year-old men exposed for 40 years at 100 dBA. We can go to ISO-1999 for distributions of ARPTS (65-year-old men) and NIPTS (40 years at 100 dBA), then simply add these decibel values to get the HTL distribution. This is really the point of our prior calculation of NIPTS using the "perfect correlation" as-

sumption: it gives us an NIPTS distribution that can be easily combined with ARPTS distributions to predict HTLs. In fact, the NIPTS distributions in ISO-1999 do not really represent the actual decibel shifts added to individuals' age-related thresholds at all. Rather they represent distributions of numbers that, when added to ARPTS values at equal percentiles, yield appropriate distributions of HTL values.

The true distribution of NIPTS (as applied to individuals) depends on the true correlation between ARPTS and NIPTS, which is certainly not 1. Later we will illustrate the distribution for the case where $r = 0$ (the truth is probably somewhere in between), but we should first note that the correlation of the two variables affects only the width of the NIPTS distribution, not its mean, because the mean of the sums of two random variables must equal the sum of the means of those variables.

HTL at the Median: Separating ARPTS and NIPTS

We can now address retrospective analysis of an individual's audiogram, in this case, a 65-year-old man with 40 years exposure at 100 dBA. To make it easy, assume his 1 kHz threshold is 20 dB HL, right at the ISO-1999 median. Common sense compels the conclusion shown in Figure 32-2: the most likely combination is simply median ARPTS (9) plus median NIPTS (11). There are an infinite number of other combinations that add up to 20 dB (e.g., 8 + 12, 7 + 13, 10 + 10, etc.), but each of these is individually less likely to occur than 9 + 11. We would conclude that 55% (11/20) of his 1 kHz loss was attributable to noise.

90th Percentile HTL

Finding the most likely combination of ARPTS and NIPTS is easy when the individual's HTL is at the population median, but that does not happen very often. We will now consider an individual with a threshold at the 90th percentile, that is, only 10% of similarly exposed workers will have higher thresholds (ISO-1999 actually uses the opposite convention and would label this the 10th percentile; the

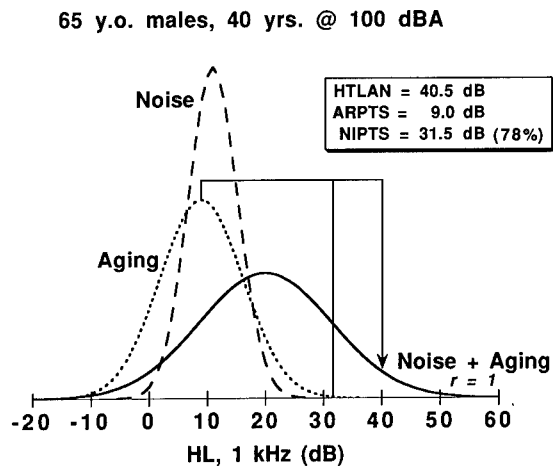


Figure 32-3 The distributions are the same as in Figure 32-2. An individual with a 90th percentile HTL (40.5 dB) is assumed to have median ARPTS, with the remainder attributed to NIPTS.

choice is arbitrary and does not affect our arguments).

Median ARPTS

Figure 32-3 illustrates one fairly common, but misguided, approach to the problem (often referred to as "age correction"). The arrowhead points to 40.5 dB, the 90th percentile point on the noise and aging distributions. Some would advocate a median ARPTS allowance (9 dB), with the remainder (31.5 dB) allocated to NIPTS. However, this implies that our subject has average susceptibility to aging, but is highly susceptible to noise.

Median NIPTS

It would be just as reasonable to assign a median NIPTS (11 dB) with the remainder (29.5 dB) allocated to ARPTS (Figure 32-4). This implies average noise susceptibility and extreme susceptibility to aging. Obviously, both median ARPTS and median NIPTS involve unwarranted assumptions about the individual unless we have data, such as an extensive audiometric record, to indicate extreme susceptibility to either noise or aging. (We can temporarily ignore another flaw in these examples: the $r = 1$ assumption in Figures 32-2–32-5 would actually *compel* ARPTS and NIPTS

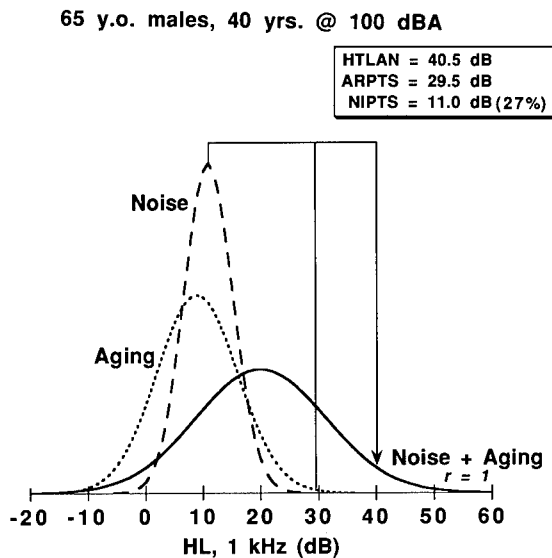


Figure 32-4 A median NIPTS allocation is illustrated for the same individual shown in Figure 32-3.

for an individual to have identical percentiles. The above arguments against median ARPTS and median NIPTS would apply to all values of $r \neq 1$).

Equal Percentile

Figure 32-3 suggested our subject's 1 kHz threshold elevation was 78% attributable to noise, while Figure 32-4 yielded a 27% noise allocation. Neither seems fair, and Figure 32-5

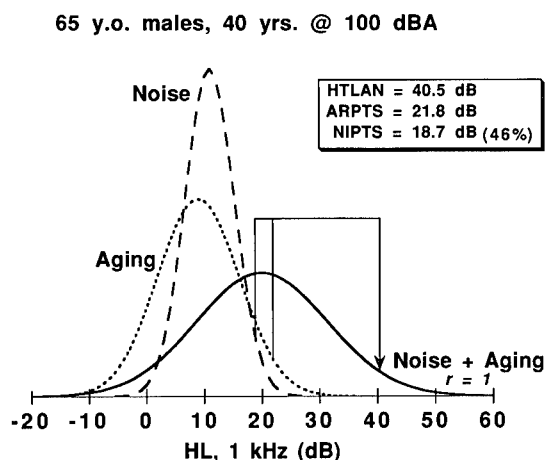


Figure 32-5 ARPTS, NIPTS, and HTL are all at the 90th percentile.

suggests a reasonable alternative: find the 90th percentile ARPTS and the 90th percentile NIPTS that (not coincidentally) add up to the 90th percentile HTL value. This approach suggests that noise is responsible for 46% of our subject's loss at 1 kHz.

Zero Correlation

Figures 32-2–32-5 were based on the ISO-1999 distributions for ARPTS, NIPTS, and HL that assume $r = 1$. However, while these NIPTS distributions are quite reasonable for the *prospective* purposes of ISO-1999, they are misleading for retrospective analysis. It is inconceivable that the susceptibilities to ARHL and NIHL are perfectly correlated, so we must examine the effects of different assumptions on our methods.

Assume ARPTS and NIPTS are uncorrelated, that is, a person's susceptibility to noise is unrelated to his or her susceptibility to age-related hearing loss. The standard deviation of the HTL distribution will be much less than the sum of the standard deviations of the ARPTS and NIPTS distributions: in fact, it is the *variances* that add up in this case. Thus, to derive the true NIPTS distribution, we find the difference between the medians of the two measured distributions (aging and noise and aging), then do the same for the variances. The result is shown in Figure 32-6. Note that the NIPTS median is the same in the $r = 0$ example as in the previous $r = 1$ examples, but that the distribution is wider.

Allocation using the $r = 0$ assumption proceeds along the same logical path as before. Median ARPTS and median NIPTS approaches are not illustrated, because they still do not make sense: each assumes the individual is average for one cause of hearing loss and extremely susceptible for the other. We have shown elsewhere³ that the $r = 0$ case can be modeled as a three-dimensional joint probability density function (JPDF), and that the most probable combination of ARPTS and NIPTS values to yield a specified HTL value is easily calculated as a function of the means and variances of the ARPTS and NIPTS distributions; this is *not* an equal-percentile solu-

65 y.o. males, 40 yrs. @ 100 dBA

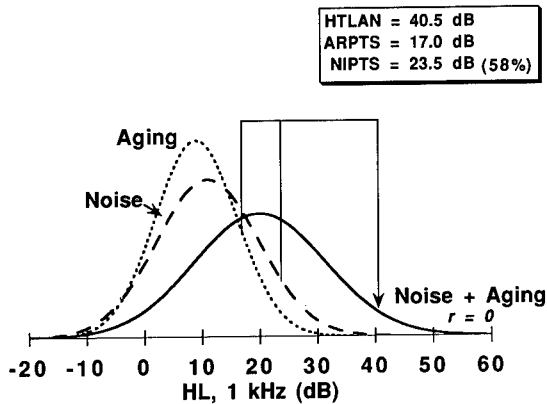


Figure 32-6 The NIPTS distribution has been recalculated assuming no correlation between ARPTS and NIPTS. For an individual with a 90th percentile HTL, the most likely ARPTS and NIPTS values are shown.

tion, except in the special case where ARPTS and NIPTS variances are identical. The results of those calculations are also shown in Figure 32-6: 17 dB ARPTS + 23.5 NIPTS = 40.5 dB HTL. NIPTS represents 58% of the total. Graphically, this set of numbers represents the peak of a bell-shaped curve⁴ formed by the intersection of the JPDP with a vertical plane representing all solutions to the equation: ARPTS + NIPTS = 40.5.

Ratio of Medians

We have found two reasonable approaches for allocating our 90th percentile HTL. If $r = 1$, use the 90th percentile ARPTS and the 90th percentile NIPTS (noise responsible for 46%). If $r = 0$, solve for the appropriate point on the JPDP (noise responsible for 58%; when HTL is above the median, the $r = 0$ method always yields a higher noise allocation than $r = 1$). If r is between 0 and 1 (partially correlated susceptibilities), as seems most likely, the best estimate would be somewhere between 46 and 58%. Now look back to Figure 32-2: the ratio of median ARPTS to median NIPTS yielded a 55% noise allocation. This is in the desired range, although somewhat closer to the $r = 0$ end. At or near median HTL, all three

methods give identical or nearly identical results. At the 90th percentile for HTL, the median ratio method is almost always intermediate between the $r = 0$ and $r = 1$ methods³; in a few cases, it yields a higher noise allocation than the $r = 0$ method, that is, it may overestimate the contribution of noise, by small amounts. Most attractively, it is much easier to compute than the other methods.

Some Audiometric Examples

HTL at Median

Figure 32-7 depicts median audiograms for a noise-exposed population (40 years at 90 dBA) and a non-noise-exposed population (60-year-old men). The regions representing ARPTS and NIPTS have been filled between 0.5 and 3 kHz (this is the range of frequencies used to estimate hearing handicap in most jurisdictions of the United States⁶). Based on our discussions so far, it is easy to estimate the relative contributions of aging and noise for an individual whose audiogram matches the noise-exposed median curve: We simply compute the median pure-tone averages (0.5, 1, 2, and 3 kHz) for ARPTS and NIPTS, and distribute the blame according to those medians (76% age, 24% noise).

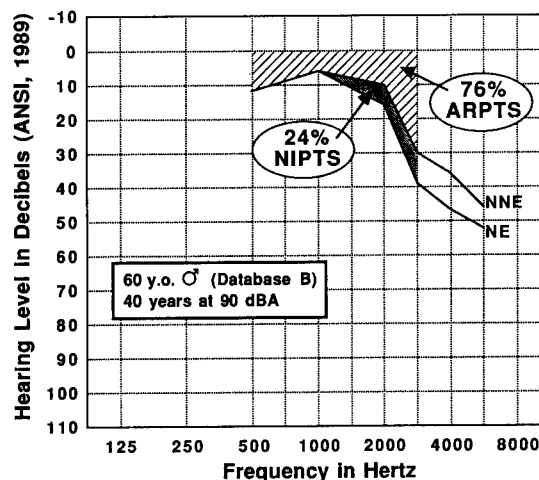


Figure 32-7 Median audiograms for noise-exposed and non-noise-exposed populations (ISO-1999). ARPTS and NIPTS contributions are shown for the 0.5–3.0 kHz range.

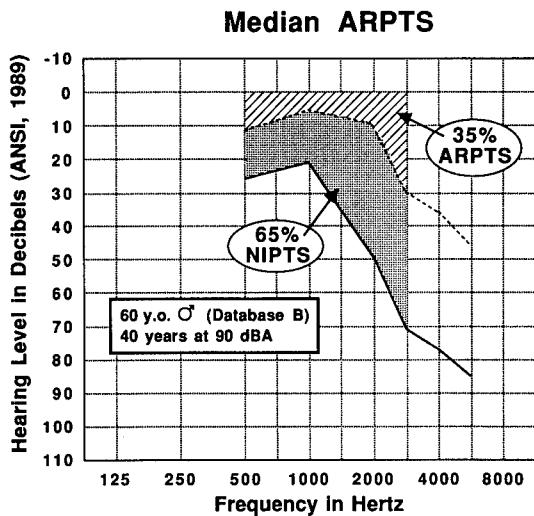


Figure 32-8 A 90th percentile audiogram is shown, along with a median-ARPTS curve, illustrating the age correction approach that assumes extreme susceptibility to noise.

90th Percentile HTL

Median ARPTS

The audiogram indicated by the solid line in Figure 32-8 represents the 90th percentile prediction from ISO-1999, that is, only 10% of a similarly exposed population would be expected to show higher thresholds at each of the frequencies plotted. The dashed line in figure 32-8 illustrates the age-correction approach; a *median* amount of ARPTS (identical to the non-noise-exposed curve in Figure 32-7) is deducted from the total, and the remainder attributed to NIPTS. For the 0.5, 1, 2, and 3 kHz pure-tone average, this results in a very different allocation from that shown in Figure 32-7 (65% NIPTS vs. 24% NIPTS).

Median NIPTS

The fallacy of the age correction approach is that it implicitly assumes the individual has average aging susceptibility and extreme susceptibility to noise. Obviously, the opposite is just as probable, and would justify the allocation shown in Figure 32-9 (we could call this noise correction). Here, a median amount of NIPTS is deducted, with the remainder

blamed on ARPTS. This results in only 11% NIPTS. The allocations in Figures 32-8 and 32-9 are both unfair, for the same reasons we rejected the allocations in figures 32-3 and 32-4. To assume median susceptibility for either ARPTS or NIPTS amounts to assuming that the individual is highly susceptible to the other.

Median Ratio

Figures 32-5 and 32-6 illustrated reasonable allocations based on the most likely combinations of ARPTS and NIPTS to yield a particular HL value. The results differ slightly, because they depend on the (unknown) degree of correlation between the susceptibilities to ARPTS and to NIPTS. Both estimates are somewhat tedious to compute, especially the $r = 0$ case. Fortunately, a much simpler estimate, based on the ratio of median ARPTS to median NIPTS, nearly always falls between the $r = 0$ and $r = 1$ allocations. This method is illustrated in Figure 32-10. The audiogram is the same one shown in Figures 32-8 and 32-9, but the allocation to ARPTS and NIPTS is based, for each frequency, on the ratio of the medians shown in Figure 32-7. The result is nearly identical to Figure 32-7: 77% ARPTS and 23% NIPTS.

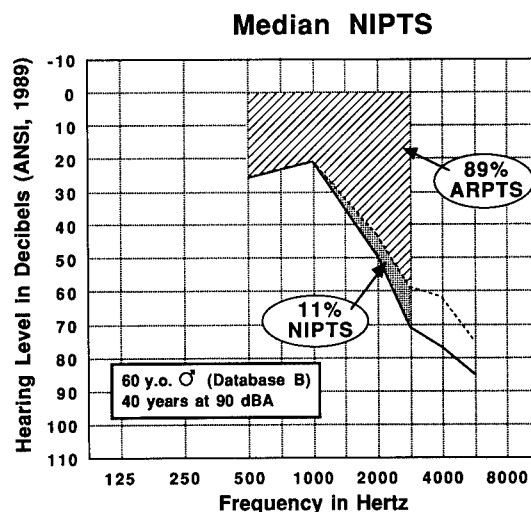


Figure 32-9 The same audiogram as in Figure 32-8 is combined with a median-NIPTS allocation.

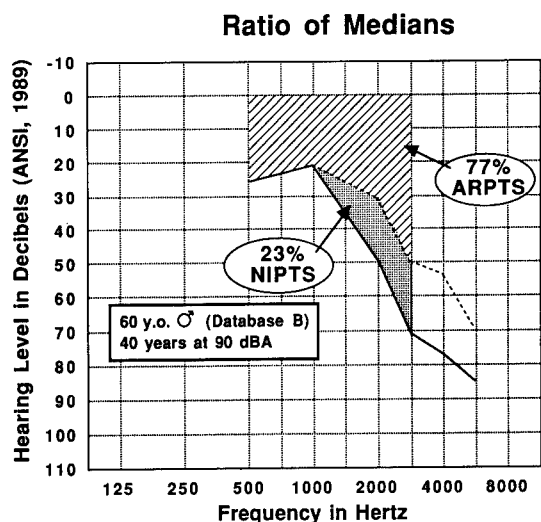


Figure 32-10 The same audiogram is again shown, but with ARPTS and NIPTS calculated based on the ratios of medians for each frequency.

Figure 32-10 shows the median ratio method applied on a frequency-by-frequency basis. However, when the audiogram is reasonable in shape, it is simpler (and only negligibly different in outcome) to compute the pure-tone averages for median ARPTS and median NIPTS, then base the allocation on the ratio between these pure-tone averages. This is the same as the Figure 32-7 allocation (24% noise). Confidence intervals can be calculated if desired.⁴

Summary

This chapter has illustrated the reasoning behind the median ratio method of allocation using ISO-1999 data. This simple method is applicable for most cases presenting claims for compensation for occupational NIHL, with the following conditions:

1. reasonably symmetrical audiograms with reasonable shape;
2. HTLs within the 5th–95th percentile range (ISO-1999) for the individual's age, gender, and exposure history;
3. no extensive audiometric record;

4. otological diagnosis of ARHL plus NIHL, without substantial contributions to HLs from other etiologies; and
5. exposure level estimate (between 85 and 100 dBA, time-weighted average) available.

Other methods are appropriate when these conditions are not met.² For example, if there is an extensive audiometric record tracking the development of an individual's hearing loss, the *trajectory* of HTL growth may show that either ARPTS (accelerating) or NIPTS (decelerating) has predominated. Methods are also available for dealing with asymmetry, extreme audiograms, changing exposure levels, etc.

The ISO-1999 model for combining ARPTS and NIPTS assumes decibel additivity (with compression), and our use of this model for retrospective allocation assumes a positive correlation between ARPTS and NIPTS susceptibilities. Discussion of these assumptions is beyond the scope of this chapter, but both are reasonable, with supporting evidence outweighing contrary evidence. No better justified models have emerged.

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Chapter 33

Compensation for Tinnitus in Noise-Induced Hearing Loss

Alf Axelsson and Ross Coles

During the last 20 years there has been a focus on many different aspects of noise-induced hearing loss (NIHL), such as pathophysiology, handicap, prevention, hearing conservation programs, hearing protection, and information to the public. In countries that have compensation for work-related conditions, the hearing loss caused by longstanding occupational noise exposure has been an increasing frequency issue. On the one hand the employers have held that the noise-exposed worker should protect himself in order to prevent hearing loss; on the other hand the workers consider it to be the employer's responsibility to create a nontraumatic work environment including a nondamaging sound climate. Many countries now have instituted compensation for NIHL.

There has also been an increasing awareness of the social wellbeing handicapping effects of hearing loss caused by noise. Furthermore, longstanding exposure to occupational noise may not only lead to hearing loss but also to tinnitus and hyperacusis. In general, tinnitus has attracted very little interest until the last 20 years. In cases with noise exposure, tinnitus occurs in approximately one-third¹ overall. Conversely, in establishing the etiology of tinnitus in a given case it turns out that about one-third of all tinnitus cases could be related to traumatic noise damage.¹ Many patients have continuous tinnitus, but do not suffer from it all the time. With or without treatment they have acquired the ability to keep the tinnitus in the back of their minds, to "forget it," to "suppress it." However, partic-

ularly in quiet environments, tinnitus may sometimes be troublesome and delay onset or return to sleep.

A patient with hearing loss does not suffer from his handicap all the time. There are usually many periods during the day when hearing is not immediately needed, that is when hearing is not "turned on." In contrast, tinnitus for most patients is continuously present, always turned on. In many cases tinnitus is far more troublesome than the hearing impairment and influences not only sleep, but also mood, concentration, and, according to the patient, often speech recognition. Consequently, it is understandable that tinnitus has become an issue concerning compensation for work-related problems.

Regulations in Different Countries

Historically there is very little published on this issue. An early German paper² addresses the questions of how to examine the patient and whether tinnitus decreases the ability to work.

In a symposium on noise, its effects, and control, there was a compilation of the regulations for different states in America.³ The authors report that 16 states gave compensation for tinnitus while 16 did not. For 12 states there was a possible compensation and for 6 states information was missing.

Glorig⁴ described a guideline used by the Veterans' Administration. This guideline lists six criteria to be met before the Veterans' Administration may consider a tinnitus com-

plaint to be valid, although of course the ultimate decision is made by a jury or a judge.

1. The complaint (or claim) that tinnitus was present and disabling must have been unsolicited. If the complaint was not present in the medical records prior to the claim, it seemed reasonable to assume that it arose as a consequence of the interview and medical history process.
2. The tinnitus must accompany a compensative level of hearing loss.
3. The treatment history must include one or more attempts to alleviate the perceived disturbance by medication, prosthetic management, or psychiatric intervention.
4. There must be evidence to support the idea of personality change or sleep disorders.
5. There must be no contributory history of substance abuse.
6. The complaint of tinnitus must be supported by statements from family or significant others.

However, it should be pointed out that these guidelines had been set for a particular compensation scheme and, like many governmental regulations, are set at a very restrictive level. We would regard them as unrealistically "tough" for more general use.

In Sweden there has been a customary increase of the NIHL compensation of 2.5–5.0% if the patient spontaneously reported troublesome tinnitus in addition to hearing loss as a consequence of longstanding occupational noise exposure. Lately, patients have claimed occupational health compensation for tinnitus only that was induced by noise or by psychic or head trauma such as whiplash injuries. Such work-related conditions may also warrant monetary compensation in Sweden.

In Australia, Macrae^{5,6} reports that there has been a general allowance for the effect of tinnitus on the ability to hear in everyday life. He has also discussed at length the possible influence of tinnitus on speech recognition, and concludes that tinnitus can affect hearing threshold levels and cause difficulties in dis-

tinguishing between pure tones and tinnitus during audiometric testing. This was also reported previously by Douek and Reid.⁷ These authors consider that tinnitus' effect on the ability to hear in everyday life must be slight, and they tend to regard broadband tinnitus as having a greater effect than narrowband or pure-tone tinnitus in this respect. The concept that tinnitus has little effect on the ability to hear in everyday life has also been supported by other investigations.⁸ These authors found that tinnitus symptoms did not contribute to the explained variance of perceived handicap in a sample of middle-aged subjects with NIHL.

In 1993 we surveyed the regulations from different countries based on an inquiry to some Western nations. The results of this inquiry are presented in Table 33-1. It can be seen that in the United States most states would not consider tinnitus in compensation for NIHL and less so than in the previous study.³ Most of the countries gave compensation for noise-induced tinnitus to some extent. In some countries there was a requirement of the presence of an NIHL determined audiometrically, but in some other countries there was no such requirement.

In Sweden a slight NIHL is not compensative in itself, but in combination with noise-induced tinnitus, may be considered compensative. Indeed, a case with normal hearing in pure-tone audiometry but with severe tinnitus that developed after an acute acoustic trauma may be compensated, as indeed it might be in the United Kingdom also, as an occupational "accident." Similarly, a whiplash injury with normal hearing and tinnitus may be compensated.

The Workup

We suggest that the workup for an alleged case of noise-induced tinnitus should be based upon:

- history, including a detailed description of the tinnitus and its effects;

Table 33-1 Tinnitus (T) Compensation Regulations

Country	Is T Considered in Compensation for NIHL	Is NIHL a Requirement?
United States*	13 states would, 26 would not 2 possible, 6 did not respond	No
United Kingdom	Yes; 0–20% increase	Yes
Canada	No	—
Australia	Yes; 5% increase	Yes
Germany	Yes; 2.5–10% increase	Yes
Denmark	Yes; 0–5% increase	Yes
Sweden	Yes; 0–20% increase	No

NIHL, noise-induced hearing loss.

* According to the American Speech and Hearing Association.

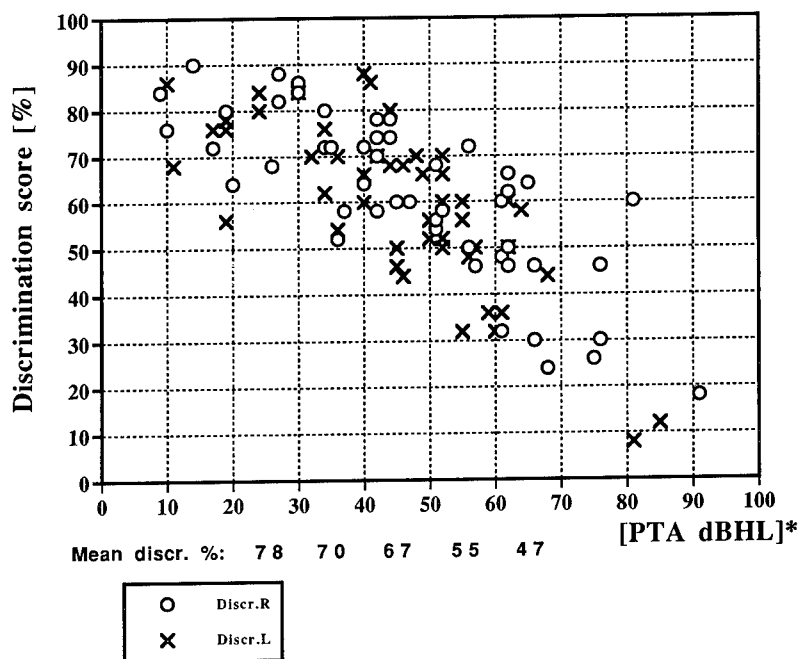
- quantitative estimation of lifetime noise exposures;
- pure-tone audiometry 250–8000 Hz;
- speech audiometry (in noise) or
- cortical electric response audiometry (ERA);
- tinnitus pitch and loudness match;
- minimum masking level of tinnitus.

Questionnaires are often used in the assessment of NIHL. It is important that such a questionnaire includes detailed information of the workplace, the type of noise, the exposure time, etc. It is also important to inquire about hearing protection and the presence of hearing conservation programs. Because tinnitus is a completely subjective symptom that cannot be confirmed or excluded, we feel that it is particularly important that the patient reports troublesome tinnitus spontaneously and not only on direct questioning. For determination of the amount of compensation it is valuable to include questions about the severity of tinnitus. We put a particular emphasis on the conditions of first appearance of tinnitus, for example, if the worker was exposed to very loud impulsive sounds with immediate appearance of tinnitus after the exposure.

It is also important to include information in the assessment about noisy leisure activities such as target practice, hunting, motor sports, exposure to loud music, and other possible

sources of tinnitus that are not work related. With leisure noise the situation is different than for occupational noise, particularly with respect to sound levels in discotheques and at pop concerts. Here there is as yet little in the way of regulation. And there is really rather little evidence of more than an occasional occurrence of disabling degrees of music-induced hearing loss. Compared to industrial noise and shooting, music-induced hearing loss is a minor problem.⁹ However, we do occasionally see cases of debilitating tinnitus arising from single exposures to high-level music where there is, as might be expected, little or no resultant hearing loss. Some of these cases of tinnitus have been observed by one of us (R.C.) to be due to "spontaneous" otoacoustic emissions. Thus, the risk of music-induced tinnitus would seem to be a more cogent argument than music-induced hearing loss for legislative limitation of the levels of amplified music to which the public may be exposed. At the very least, there would seem to be a strong case for epidemiological and clinical research on music-induced tinnitus coupled with prospective studies of the effects of high-level music exposures on spontaneous otoacoustic emissions and post-exposure tinnitus.

The assessment also includes pure-tone audiometry and either speech audiometry in noise or CERA in order to check the subjects



Median discrimination score 60%

*Pure tone average 3,4,6,8 kHz

Figure 33-1 Speech audiometry S/N +10 dB (100 ears). Subjects without tinnitus.

"auditory honesty,"¹⁰ and speech audiometry, possibly in noise. Further, regular tinnitus examinations including tinnitus pitch and loudness matching as well as the maskability of tinnitus should be included in the assessment.

In investigations in Gothenburg the first author has not been able to show any influence of tinnitus on speech recognition in noise in a sample of patients with and without tinnitus (Figures 33-1, 33-2). These figures show almost identical results that demonstrates that, at least during the test situation, there is no influence of tinnitus on speech recognition in noise.

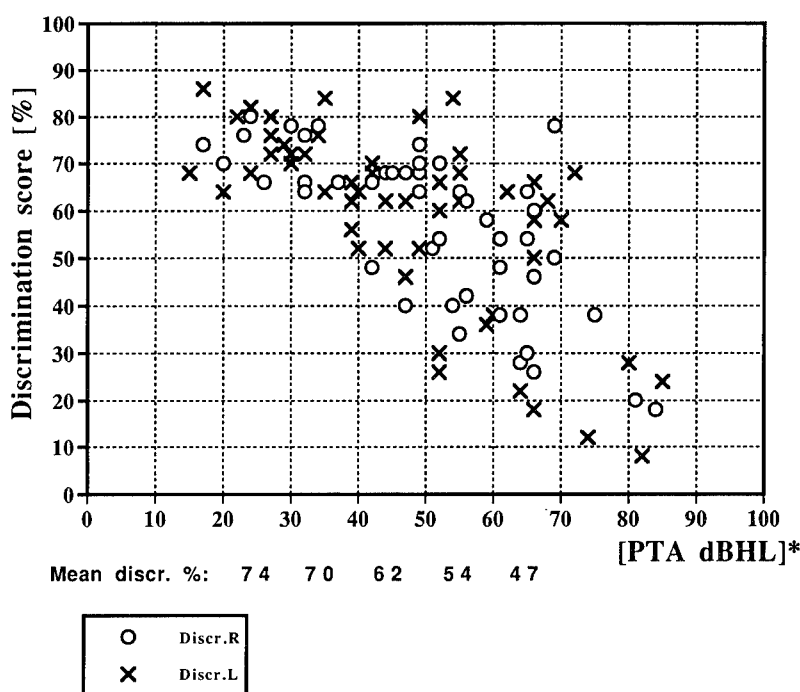
It may also be helpful to carry out repeated examinations in cases where there is reason to doubt the tinnitus information. This should include a new workup and tinnitus analysis. In cases where the patient can reproduce fairly similar test results, we can adopt the term

"audiological reliability." A patient who exaggerates his symptoms often has great difficulties in repeating the falsely pretended results on a later occasion. However, we must also bear in mind that tinnitus may vary in quality and loudness over time, that tinnitus matching is often a difficult task, and that some people are simply not good at audiometry.

Requirements

On the basis of the history and the audiological examinations the following requirements could be made in evaluating tinnitus for compensation:

1. Spontaneous reports about tinnitus can be regarded as stronger evidence than a positive answer in a questionnaire. The information should seem reasonably like that of a case



Median discrimination score 64%

*Pure tone average 3,4,6,8 kHz

Figure 33-2 Speech audiometry S/N +10 dB (100 ears). Subjects with tinnitus.

of troublesome noise-induced tinnitus. What is the main complaint of the plaintiff, hearing loss or tinnitus?

2. The symptom of tinnitus should be clearly diagnosable as originating, at least in major part, from occupational noise exposure, not leisure activities or other causes. In case of both occupational and leisure noise a quantitative determination should be made of the noise exposures. In most cases it is reasonable to assume that the occupational noise exposure is much greater than from leisure activities.

3. In cases where the tinnitus is noise induced the configuration of the pure-tone audiogram should be typical or compatible with NIHL. In case of a history of acute acoustic trauma the audiometric configuration may be asymmetrical. In cases where the tinnitus was induced by head trauma, whiplash injury, or

trauma against the external ear or tympanic membrane, the pure-tone audiogram may show different configurations.

4. In cases of noise-induced tinnitus, the tinnitus analysis would most commonly demonstrate a high-pitched tinnitus.

5. There should be an acceptable correlation between the pure-tone audiogram and the results of speech (in noise) audiometry or CERA, to check the subjects audiological honesty.

6. In doubtful cases, where reexamination is made, there is a desirability of test-retest consistency in the pure-tone audiograms, speech recognition tests, and tinnitus analyses.

7. We have seen several cases where tinnitus was induced by psychic trauma with occupational origin, for instance by mobbing,

threat of getting fired, burn-outs, fear of economical problems, etc. In these cases we have accepted a relationship to occupational factors provided they are prominent and that there is no other more likely explanation for the tinnitus. Most of these cases have had normal hearing and bilateral tinnitus.

Discussion

It is apparent that there are many problems with compensation for tinnitus. First, the symptom is completely subjective. Yet, we maintain that a thorough workup with a detailed history and audiometric analysis in most cases at least gives an impression of truthfulness.

In the literature there is often a discussion about what should be compensated for in cases of tinnitus, and mostly there seems to be an issue that compensation should be paid for decreased speech perception rather than for the general annoyance or suffering associated with tinnitus. As shown in our own investigations, we could not find any influence of tinnitus on speech recognition, at least not in the test situation.

An obvious problem is the occupational etiology in some cases, however, without exposure to noise. Similarly to noise-induced occupational tinnitus, we feel that physical influences such as whiplash injuries and head and ear trauma could be compared to noise-induced tinnitus except for the audiometric findings. However, in cases of psychic trauma the relationship is much more difficult and most cases would probably be denied compensation.

Another problem is for what the compensation is paid. The patients may for instance maintain that they may have an alteration in their insurance premiums because of their symptoms, and others may consider the fact that they are less attractive on the labor market, provided that the new employer is informed about the tinnitus and hearing loss. Our own attitude is that it is mainly the annoyance and often resultant depression and decreased quality of life that should be compensated.

It can of course be argued that economic compensation cannot decrease the psychological impact of tinnitus on the patient, but of course neither can it decrease the disability resulting from virtually any other form of occupational injury or disease, including NIHL.

Conclusion

1. A common cause of tinnitus is NIHL.
2. Of all cases with a history of noise exposure, about one-third have tinnitus.
3. If the attitude is not to compensate or to reduce the compensation where there has been contributory negligence in lack of use of hearing protection, tinnitus is not different from hearing loss.
4. Permanent severe tinnitus diminishes the quality of life, at least as much as hearing loss.
5. Consequently, if compensation is given for hearing loss it should also be given for tinnitus. While tinnitus is important as another handicapping end product of occupational noise exposure, it is unlikely to be accepted as a reason for increased effort or tighter regulation on industrial hearing conservation. This is because the criteria for its prevention are the same as for NIHL and regulations and procedures to prevent the latter are already well established. Its occurrence in those exposed to high levels of amplified music may, however, become an important argument for control of music levels.
6. The subjective character of tinnitus necessitates a careful history and workup before compensation.

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Chapter 34

Current Standards for Occupational Exposure to Noise

Alice H. Suter

Terminology

Before initiating a discussion of current standards for occupational exposure to noise it would be useful to examine some of the language used by standards-making officials.

The terms "regulation," "standard," and "legislation" are often used interchangeably, although they have slightly different meanings, at least in the United States. A regulation is a rule or order prescribed by an authority (usually the government) and is usually more formal than a standard. Regulations often incorporate standards. A standard is a codified set of rules or guidelines, much like a regulation, but can be developed under the auspices of a consensus group, such as the American National Standards Institute (ANSI). Legislation consists of laws prescribed by authorities (lawmakers), in the United States by the Congress or by local governing bodies. These laws often enable government bodies to promulgate regulations, such as the regulations issued by the Occupational Safety and Health Administration (OSHA) and the Mine Safety and Health Administration (MSHA) in the US Department of Labor.

Elsewhere in the world, many national standards are called "legislation." Some official bodies use the terms standards and regulations as well. The Council of the European Communities (CEC) uses the term "directives." All members of the European Community needed to "harmonize" their standards (regulations or legislation) with the 1986 CEC Directive on occupational noise exposure by

the year 1990.¹ This means that the standards and regulations of the member countries had to be at least as protective as the CEC Directive.

Some nations use a "code of practice," which is somewhat less formal than the terms discussed above. For example, the Australian national standard consists of two short, mandatory paragraphs, followed by a 35 page code of practice that provides practical guidance on how the standard should be implemented. Codes of practice usually do not have the legal force of regulations or legislation. Another term that is used occasionally is "recommendation," which is more like a guideline than a mandatory rule. Recommendations are not enforceable, but they do provide guidance.

Basis for Damage-Risk Criteria

Many factors enter into the development of noise standards in addition to data describing the amount of hearing loss resulting from a certain amount of noise exposure. There are both technical and policy considerations.

What Is Acceptable Risk?

A good example of a nontechnical policy consideration is the question of what proportion of the population should be protected. In other words, what is an acceptable amount of risk in the exposed population?

In earlier years, regulatory decisions were made that allowed considerable amounts of hearing loss from noise exposure. Table 34-1

Table 34-1 Risk of Exceeding a 25 dB Fence at 500, 1000, and 2000 Hz After 40 Years Exposure

Organization	Noise Exposure in A-Weighted SPL (dB)	Percent Risk
ISO	90	21
	85	10
	80	0
EPA	90	22
	85	12
	80	5
NIOSH	90	29
	85	15
	80	3

From OSHA, 1981.

shows the percentage of the exposed population that was expected to incur a hearing handicap (defined then as an average hearing threshold greater than 25 dB at 500, 1000, and 2000 Hz) as a function of noise exposure level. The Department of Labor promulgated the first US civilian standard for occupational noise exposure knowing that more than 20% of the population exposed to the 90 dB(a) permissible exposure limit (PEL) would suffer a "disabling loss of hearing."² The acceptability of this amount of risk is certainly open to question.

How Much Hearing Should Be Saved?

Another important consideration is what are we protecting people against? Should we protect even the most sensitive members of the exposed population against any loss of hearing? This was the US Environmental Protection Agency's mandate from Congress in the Noise Control Act of 1972. Or should we protect against a compensable hearing handicap? It amounts to a question of which hearing loss formula to use, and different governmental bodies have varied widely in their selections. The Occupational Safety and Health Administration (OSHA) in the US Department of Labor is committed to the term "material impairment" by its legislative mandate. OSHA has defined "material impairment of hearing" as

an average hearing threshold level of 25 dB at the frequencies 1000, 2000, and 3000 Hz. Using this definition, the risk would be somewhat larger than it would be using the 500, 1000, and 2000 Hz combination.

If there is to be no risk from noise exposure, the PEL would have to be as low as 75–80 dB(A). In fact, the CEC has established an equivalent level (L_{eq}) of 75 dB(A) as the level at which the risk is negligible, and this level has also been put forward as a goal for Swedish production facilities.³ In general, the prevailing thought on this subject is that it is acceptable for a noise-exposed workforce to lose some hearing, but not too much. As for how much is too much, there is no consensus.

NIPTS Versus Risk

There are two ways to present criteria for noise-induced hearing loss: noise-induced permanent threshold shift (NIPTS) and percentage risk. NIPTS is the amount of permanent threshold shift (PTS) remaining in a population after subtracting the threshold shift that would occur "normally" from causes other than occupational noise. The percentage risk is the percentage of a population developing a certain amount of hearing loss (often referred to as "crossing a fence") due to noise after subtracting the percentage of a similar but non-noise-exposed population that would cross that same fence due to aging and other causes. This concept is sometimes called "excess risk." Neither method is without problems.

The problem with using NIPTS is that it is difficult to summarize the effects of noise on hearing. The data are usually set out in a large table showing noise-induced threshold shift for each audiometric frequency as a function of noise level, years of exposure, and population centile (arranged according to sensitivity to noise). The concept of percent risk is more attractive because it uses single numbers and appears easy to understand. But the trouble is that the risk can vary enormously depending on a number of factors, particularly the height of the hearing threshold level fence and the frequencies used to define hearing impairment (or handicap). In most circumstances,

Table 34-2 Example of Population at Risk From Noise Exposure

Noise exposure level: 90 dB (A)
 Gender: male
 Age: 50
 Years of exposure: 30
 Nonnoise population: Database A
 Frequencies: av. 1, 2, 4 kHz
 Fence: 27 dB
 Risk from aging: 11.5%
 Risk from noise + aging: 18%
 Percent risk: 6.5%

From ISO 1999, Annex D.⁴

using a relatively low fence and high frequencies in the definition will increase the apparent risk of a given level of noise exposure. Likewise, raising the fence and lowering the audiometric frequencies will result in a decreased risk.

With both methods the user needs to be sure that the exposed and nonexposed populations are carefully matched for such factors as age and nonoccupational noise exposure.

Table 34-2 gives an example of the percentage risk method taken from ISO 1999, Annex D.⁴ According to the ISO data and method, the risk in a male population due to noise at 90 dB(A) after 30 years of exposure is 6.5% using the frequencies 1000, 2000, and 4000 Hz and a fence of 27 dB.

If the same data and method were used, but changing just a few parameters, the risk becomes 15%, as in Table 34-3. By examining hearing impairment at age 60 with 40 years of exposure and by lowering the fence by 2 dB, the risk has been increased considerably. Moreover, the risk from aging alone has increased from 11.5% to 30% in these 10 years.

If one should ask whether Table 34-2 or 34-3 is correct, the answer is that they both are, and that it is up to the policymakers to decide the level of the fence and whether to use a duration of 30 or 40 years. But comparing these risk figures to those in Table 34-1, one can see that there are some fairly large discrepancies among the older and newer methods of calculating percentage risk. The fact that the older

Table 34-3 Example of Population at Risk From Noise Exposure

Noise exposure level: 90 dB (A)
 Gender: male
 Age: 60
 Years of exposure: 40
 Nonnoise population: Database A*
 Frequencies: av. 1, 2, 4 kHz
 Fence: 25 dB
 Risk from aging: 30%
 Risk from noise + aging: 45%
 Percent risk: 15%

ISO 1999 data and method used.⁴

*In this case the calculation includes a small adjustment to correct for the fact that the standard has normalized hearing threshold levels of 20-year-old men to 0 dB. Without this adjustment the percentage risk would be approximately 3% higher.

ISO standard, along with the EPA and NIOSH methods, yields risk figures that are considerably higher than the newer standard, even though the newer standard uses much the same data, is somewhat mysterious.

Decisions

It would be quite surprising if every decision-making body considered the appropriate frequencies, fences, years of exposure, non-noise-exposed control population, and the number of people to be protected, as it went through the NIPTS and percentage risk process. OSHA did consider most of these factors when drafting its 1991 amendment to its noise standard for hearing conservation programs, but the agency never determined the percentage of the exposed population that should be protected. By setting the action level at 85 dB(A), OSHA tacitly assumed that a risk of between 10 and 15% was acceptable.

In the introduction to the noise standard, those who drafted ISO 1999 state:

The selection of maximum tolerable or maximum permissible noise exposures, and protection requirements as well as the selection of specific formulae for handicap risk assessment or compensation purposes, require consideration of ethical, social, economic and political

factors not amenable to international standardization. Individual countries differ in their interpretation of these factors and these factors are therefore considered outside the scope of this International Standard.

In all probability, most nations draft standards and regulations that attempt to keep the risk at a minimum level, while taking technical and economic feasibility into account, but without coming to consensus on such matters as the frequencies, fence, or non-occupationally exposed comparison group.

Nations' Standards

Table 34-4 summarizes some of the main features of the noise exposure standards of several nations. Most of the information is current as of this publication, but there may be some data that have been recently revised. The reader is advised to consult the newest versions of the individual nations' standards.

Trends

At this time, most nations use 85 dB(A) as the PEL. A few use 90 dB(A), some require varying PELs depending upon the nature of the work (China, Germany, and Norway), and one nation, The Netherlands, has a PEL of 80 dB(A). There seems to be general agreement that everyone loses some hearing when exposed to a PEL above 90 dB(A), some lose hearing above 85 dB(A), a few above 80 dB(A), and that the safe level is somewhere around 75–80 dB(A).

Even though most nations have placed the PEL at 85 dB(A), about half still use 90 dB(A) for compliance with requirements for engineering control, as allowed by the CEC. Some, such as Germany and Australia, as well as the CEC, urge employers to control noise to the lowest level practicable.

Nearly every nation listed above has adopted the 3 dB exchange rate, with the exception of Brazil, Israel, and civilian United States, which use the 5 dB rule. The US Army and Air Force now use 3 dB.

Most nations limit impulsive noise exposure to 140 dB peak sound pressure level

(SPL), with the exception of Brazil at 130 and France at 135 dB. Quite a few limit continuous noise to 115 dB(A), and Norway requires a maximum of 110 dB(A).

In general, there is no indication of the degree to which these standards and regulations are enforced. Some specify that employers "should" take certain actions (as in codes of practice or guidelines), while most specify that employers "shall." Standards that use "shall" are more apt to be mandatory, but individual nations vary widely in their ability and inclination to enforce. Even within the same nation, such as the United States, enforcement of occupational noise standards may vary considerably with the administration in power.

Features of Nations' Noise Standards

In addition to protecting workers against hearing loss, several nations include requirements to protect against other adverse effects of noise. Also, unlike most noise standards in the United States, many other nations have noise standards for specific workplaces, processes, and equipment or machinery.

Standards to Protect Against Effects Other Than Hearing Loss

Both the CEC directive and the German standard acknowledge that workplace noise involves a risk for the health and safety of workers beyond hearing loss, but that current scientific knowledge of the nonauditory effects does not enable precise safe levels to be set. They assume that reduction of noise will lower the risks of these effects.

The nonauditory health effects are not the only ones addressed. For example, the Norwegian standard includes a requirement that noise levels must not exceed 70 dB(A) in work settings where speech communication is necessary. The German standard advocates noise reduction for the prevention of accident risks, and both Norway and Germany require a maximum noise level of 55 dB(A) to enhance concentration and prevent stress during mental tasks.

Table 34-4 Permissible Exposure Limits, Exchange Rates, and Other Requirements for Noise Exposure According to Country

Country, Date	PEL $L_{av.}$ [8h dB (A)]	Exchange Rate [dB (A)]	L_{max} rms L_{peak} SPL	Level* dB (A)		Comments
				Eng. Control	Audio. Test	
Australia, 1993	85	3	140 dB peak	85	85	Note 1
Brazil, 1992	85	5	115 dB (a) 130 dB peak	85		
Canada, 1990	87	3		87	84	Note 2
CEC, 1986	85	3	140 dB peak	90	85	Note 3 Note 4 Note 5
China, 1985	70-90	3	115 dB (a)			
Finland, 1982	85	3		85		
France, 1990	85	3	135 dB peak		85	
Germany, 1990	85, 55, 70	3	140 dB peak	90	85	Note 3 Note 6
Hungary	85	3	140 dB peak 125 dB (a)	90		
India, 1989	90		115 dB (a) 140 dB (a)			Note 7
Israel, 1984	85	5	115 dB (a) 140 dB peak			
Italy, 1990	85	3	140 dB peak	90	85	
Netherlands, 1987	80	3	140 dB peak	85		Note 8
New Zealand, 1981	85	3	115 dB (a) 140 dB peak			Note 9
Norway, 1982	85, 55, 70	3	110 dB (a)		80	Note 10
Spain, 1989	85	3	140 dB peak	90	80	
Sweden, 1992	85	3	115 dB (a) 140 dB (c)	85	85	
United Kingdom, 1989	85	3	140 dB peak	90	85	
United States, 1983	90	5	115 dB (a) 140 dB peak	90	85	Note 11

Sources: Pamela Gunn, Dept. Occup. Health, Safety & Welfare, Perth, Western Australia (personal communication); Tony F.W. Embleton, Nobleton, Ontario, Canada (personal communication); ILO Noise Regulations and Standards, CIS data base, International Labour Office, Geneva, Switzerland; and publications of various nations.

*PEL = Permissible exposure limit

**Like the PEL, the levels initiating the requirements for engineering controls and audiometric testing also, presumably, are average levels.

¹Levels for engineering controls, hearing tests, and other elements of the hearing conservation program are defined in a code of practice.

²There is some variation among the individual Canadian provinces: Ontario, Quebec, and New Brunswick use 90 dB(A) with a 5-dB exchange rate; Alberta, Nova Scotia, and Newfoundland use 85 dB(A) with a 5-dB exchange rate; and British Columbia uses 90 dB(A) with a 3-dB exchange rate. All require engineering controls to the level of the PEL. Manitoba requires certain hearing conservation practices above 80 dB(A), hearing protectors and training on request above 85 dB(A), and engineering controls above 90 dB(A).

³The Council of the European Communities (86/188/EEC) and Germany (UVV Lärm-1990) state that it is not possible to give a precise limit for the elimination of hearing hazard and the risk of other health impairments from noise. Therefore the employer is obliged to reduce the noise level as far as possible, taking technical progress and the availability of control measures into account. Other EC nations may have adopted this approach as well.

⁴Those countries comprising the European Community were required to have standards that at least conformed to the CEC Directive by January 1, 1990.

Table 34-4 (Continued)

- ⁵China requires different levels for different activities: eg. 70 dB(A) for precision assembly lines, processing workshops, and computer rooms; 75 dB(A) for duty, observation, and rest rooms; 85 dB(A) for new workshops; and 90 dB(A) for existing workshops.
- ⁶Germany also has noise standards of 55 dB(A) for mentally stressful tasks and 70 dB(A) for mechanized office work.
- ⁷Recommendation.
- ⁸The Netherlands' noise legislation requires engineering noise control at 85 dB(A) "unless this cannot be reasonably demanded." Hearing protection must be provided above 80 dB(A) and workers are required to wear it at levels above 90 dB(A).
- ⁹New Zealand requires a maximum of 82 dB(A) for a 16-hour exposure. Ear muffs must be worn in noise levels exceeding 115 dB(A).
- ¹⁰Norway requires a PEL of 55 dB(A) for work requiring a large amount of mental concentration, 85 dB(A) for work requiring verbal communication or great accuracy and attention, and 85 dB(A) for other noisy work settings. Recommended limits are 10 dB lower. Workers exposed to noise levels greater than 85 dB(A) should wear hearing protectors.
- ¹¹These levels apply to the OSHA noise standard, covering workers in general industry and maritime. The US military services require standards that are somewhat more stringent. The US Air Force and the US Army both use an 85-dB(A) PEL and a 3-dB exchange rate.

Standards for Various Workplaces, Processes, and Equipment

Some countries have special noise standards for different kinds of workplaces, while others include noise as one of many regulated hazards in a particular process. The latter type are called "vertical" standards in the United States. Still other standards apply to specific types of equipment or machines. Tables 34-5-34-7 give examples of these kinds of stan-

dards. These tables do not reflect exhaustive surveys, but are meant to illustrate the types of noise regulations that various nations use.

In addition to these standards, some nations have promulgated separate standards for hearing protection devices (such as the CEC, The Netherlands, and Norway) and for hearing conservation programs (such as France, Norway, Spain, Sweden, and the United States).

Some nations use innovative approaches to attack the occupational noise problem. For example, The Netherlands has a separate standard for newly constructed workplaces, and Australia and Norway give information to employers for instructing manufacturers to provide quieter equipment.

Table 34-5 Workplaces Regulated for Noise

Workplace	Nation
Auto repair shops	Norway
Fish oil and meal factories	Norway
Foundries	Norway
Motor vehicle cabs	Finland
	United States
Nursery schools and day care centers	Sweden
Observation and rest areas	China
Offices	Germany
	Japan
	Netherlands
Restaurants	Norway
	Sweden
Shipboard	Finland
Steel mills	Sweden

Table 34-6 Work Processes Regulated for Noise

Work Process	Nation
Computer work	Sweden
Cutting and welding	Denmark
Excavation (construction)	Sweden
Fish canning and processing	Denmark
Precision assembly lines	China

Table 34-7 Equipment and Machinery Regulated for Noise

Equipment or Machine*	Nation
Air compressors	CEC
Chain saws	France
	Japan
	Norway
	Sweden
Construction equipment	CEC
	Finland
	France
Snow removal equipment	Finland
	Switzerland
Tractors	Finland
	Hungary
	Netherlands
Woodworking machines	China
	Netherlands
	Norway
	Sweden
	United Kingdom

*In 1989 the CEC established a directive requiring manufacturers to include instructions on noise level when any machinery exceeded 70 dB (A) or 130 dB (C), or when sound power levels exceeded 85 dB (A).¹

Draft Recommendations of the International Institute of Noise Control Engineering

In 1992, the General Assembly of the International Institute of Noise Control Engineering (I-INCE) approved a Working Party to review current knowledge and practice on upper noise limits in the workplace. Made up of 11 members from 8 different countries, the Working Party is Chaired by Tony F.W. Embleton. The group's preliminary recommendations are summarized in Table 34-8.

These recommendations appear to reflect the consensus of the international community at this time, as well as summarize the present situation of international noise standards and regulations for occupational exposure to noise.

References

1. Council of the European Communities. *Council Directive of 12 May 1986 on the Protection of Workers*

Table 34-8 Draft Recommendations of I-INCE Working Party on Upper Noise Limits in Workplace

Exposure limit:	8h L_{eq} of 85 dB (A)
Impulse noise:	140 dB unweighted peak or 130 dB "impulse"
Exchange rate:	3 dB regardless of time variation
Engineering controls:	Building design should provide for sound and vibration isolation between noisier and quieter areas
	Purchase specifications for new and replacement machinery should contain clauses specifying maximum allowable sound levels
Audiometric tests:	Preplacement hearing tests $> L_{eq}$ 80–85 dB and subsequent tests at intervals that depend on exposure levels

From Embleton.⁷

From the Risks Related to Exposure to Noise at Work (86/188/EEC);1986.

2. US Dept. Labor. *Guidelines to the Department of Labor's Occupational Noise Standards for Federal Supply Contracts*. Bulletin 334. Washington, DC: U.S. Dept. Labor, Workplace Standards Administration, Bureau of Labor Standards; 1970.

3. Kihlman T. Sweden's action plan against noise. *Noise/News Int* 1993;1:194–208.

4. International Organization for Standardization. *ISO 1999: Acoustics: Determination of occupational noise exposure and estimate of noise-induced hearing impairment*. Geneva: 1990.

5. Occupational Safety and Health Admin. Occupational noise exposure: Hearing conservation amendment. *46 Fed. Reg.* 1981:4078–4179.

6. Council of the European Communities. *Council Directive of 14 June 1989 on the Approximation of the Laws for the Member States Relating to Machinery (89/392/EEC);1989.*

7. Embleton TFW. Report by I-INCE Working Party on Upper noise limits in the workplace. *Proceedings of INTER-NOISE 94*. Yokohama, Japan; 1994.

Chapter 35

Ototoxic Effects of Chemicals Alone or in Concert With Noise: A Review of Human Studies

John R. Franks and Thais C. Morata

Occupational hearing loss is one of the 10 leading work-related disorders in the United States as in many other countries. At least one million workers in manufacturing have sustained job-related hearing impairment (defined as greater than a 25 dB average threshold hearing level at 1, 2, and 3 kHz), and about half a million of these have moderate to severe hearing impairment (defined as greater than or equal to a 40 dB average threshold hearing level at 1, 2, and 3 kHz).¹ In addition to the issue of the impact of permanent hearing loss, the cost of this hearing loss is substantial. Workers file compensation claims for hearing losses thought to result from occupational noise exposure, and the cost of these claims for the period of 1977–1987 was estimated at \$800 million.² Even within the American Federal Government, between 1983 and 1992, over \$320 million was paid to civilians in compensation for occupational hearing loss.³

Attention in occupational health to the prevention of hearing loss has focused almost entirely on workplace noise, and the literature on the effects of noise exposure on hearing is extensive. Conversely, the effects of other factors such as medical conditions, vibration, physical work load, and chemical exposure have received comparatively little attention.^{4–7}

Still, it is not yet commonly recognized that various chemicals can also be ototoxicants on

their own. The ototoxicity of environmental agents such as metals, solvents, and asphyxiants, and their interaction with noise, are issues just beginning to receive attention in the international literature.^{8–10} Additional studies on this topic are necessary because there is evidence that occupational hearing loss may be caused not only by noise but also by other factors in the work environment. Because some industrial chemicals are known to be ototoxic, it is plausible to expect that if these chemicals occurred in high enough concentrations in the workplace, they could affect hearing. Currently, occupational legislation does not consider environmental chemicals hazardous to hearing. Thus, there may be numerous workers with unmet needs concerning hearing conservation.

Investigation of the combined effects of noise and other agents was recommended in the NIOSH Proposed National Strategy for the Prevention of Noise-Induced Hearing Loss under Prevention Strategy—Research that states: “Determine through investigations the degree to which noise interacts with other agents in the work environment (solvents, metals, prescription drugs, etc.) to affect hearing” (III.C.2.d., pg 9).¹¹

The major focus of this chapter is the human data on the effects of environmental and industrial chemicals on hearing and the potential for an ototraumatic interaction with noise.

Effects of Environmental and Industrial Chemicals on Auditory System

The auditory disorders associated with medicinal drugs such as loop diuretics, aminoglycosides, salicylates, antitumor agents, and quinine and its derivatives have been well publicized.⁵ Conversely, the ototoxic/vestibulotoxic effects of environmental and industrial chemicals have received comparatively little attention. Metals, chemical asphyxiants, and solvents are three classes of chemical compounds that have been studied following case reports linking accidental overexposure or substance abuse of these compounds with neurotoxic and/or ototoxic effects.¹²

There has been a growing interest in the effects that simultaneous exposure to noise and chemicals might have on hearing.^{13–19} In a review paper that briefly discussed five occupational studies and four case reports, an ototraumatic interaction between noise and organic solvents was suggested and its biological plausibility discussed.¹³ It has been observed that the incidence of sensorineural hearing loss was higher than expected in workers exposed to solvents. In addition, organic solvents are well known for their neurotoxic effects that can give rise in exposed workers to both central and peripheral nervous system injuries. The principal neurological syndrome affecting the peripheral nervous system is the occurrence of axonal degeneration with attendant symptoms and signs of peripheral neuropathy. It has been argued that solvents could injure the sensory cells and peripheral endings of the cochlea. It was hypothesized that, because solvent-related effects have been found in the brain, a retrocochlear influence on hearing could also be expected.¹³

In a 20 year longitudinal study of hearing sensitivity in 319 employees from different departments of industry, a remarkably large proportion of the workers in the chemical sector showed pronounced hearing loss (23%) as compared to groups from nonchemical environments (5–8%).¹⁴ This effect was found despite the lower noise levels in the chemical

department (80–90 dBA) when compared to other divisions (95–100 dBA). Thus, the exposure to industrial solvents (not identified in the article) was implicated as an additional causative factor for those hearing losses.

Animal Evidence

The ototoxicity of solvents, metals, and asphyxiants has been investigated and, in some cases, demonstrated in animal experiments.⁸ In rats and mice, a toluene-induced sensorineural hearing loss was observed, despite different modes of administration.^{9,18,20,21} There is strong evidence for a cochlear site of the damage caused by toluene indicating that ototoxicity, and not neurotoxicity, is most likely the process by which toluene affects the auditory system.⁹ Both xylene and styrene have also been shown to affect the auditory system and both seem to have a more potent ototoxic effect than toluene.^{16,20} Noise exposure has been shown to interact synergistically with previous toluene exposure, with the sequence of exposure affecting the outcome.^{9,22} Exposure to mixed solvents also generates interactive effects on the auditory system. Auditory brainstem responses (ABRs) have indicated an additive loss of auditory sensitivity in rats after exposure to styrene and trichloroethylene.²³ *n*-Hexane has not been shown to cause any permanent hearing impairment in rats. However, when rats were exposed to a mixture of toluene and *n*-hexane a potentiation of the effects of toluene took place, whereas an antagonistic effect was observed on the peripheral nerve conduction velocity.^{24,25} Similar results were obtained in rats exposed to *n*-hexane mixed with xylene.²⁶

Human Evidence

Solvents

The studies conducted examining the effect of exposure to solvents on hearing sensitivity in humans are displayed in Table 35-1. Studies on the effects of voluntary inhalation of the

Table 35-1 References on Effects of Exposure to Solvents on Human Auditory System (studies on effects of voluntary inhalation of solvents not included).

Toxicant	Effect	Reference
<i>n</i> -Butanol	Prevalence of HL higher for <i>n</i> -butanol exposed than for noise exposed	Velazquez et al. ²⁷
TCE	Abnormal auditory and vestibular functions	Szulc-Kuberska et al. ²⁸
	Prevalence of HL higher for TCE exposed children than for nonexposed	ATSDR ²⁹
Carbon disulfide (CS ₂)	Prevalence of HL higher for CS ₂ plus noise exposed than for noise-exposed workers; vestibular disorders	Sulkowski ³⁰ Morata ¹⁹
<i>n</i> -Hexane	Prolonged Waves V and I-V, III-V IPL Prolonged Waves V and I-V IPL	Hirata et al. ³¹ Chang ³² Huang and Chu ³³
Styrene	Significant differences in thresholds among least and most exposed Central auditory and vestibular disorders Styrene enhanced effects of noise in specific frequencies (approached significance)	Muijsers et al. ³⁴ Möller et al. ³⁵ Sass-Kortsak ³⁶
Toluene	Prevalence of HL higher for toluene plus noise and solvents only than for noise-exposed workers; vestibular disorders	Morata ¹² Morata et al. ³⁷
Solvent mixtures	Prolonged Waves I and I-III, I-V, III-V IPL Central auditory and vestibular disorders	Abbate et al. ³⁸ Ödkvist et al. ³⁹ Möller et al. ⁴⁰
	High prevalence of HL Prevalence of HL higher for solvents plus noise exposed than for noise and nonexposed workers	Bielski ⁴¹ Taniuchi et al. ⁴²
	Adjusted relative risk for HL higher for solvents exposed than for nonexposed workers, lower than risk for noise exposed	Jacobsen et al. ¹⁰

HL, hearing loss; TCE, trichloroethylene; CS₂, carbon disulfide; IPL, interpeak latency.

chemicals were not included because the exposure patterns are very different from those in industrial workplaces.

n-Butanol

In a study conducted at a small cellulose acetate ribbon factory in Mexico, 11 workers were examined.²⁷ They were exposed to levels around 80 parts per million (ppm) of *n*-butanol and noise levels of 75 dBA. An increased prevalence of hearing loss was observed for these workers (9 of 11) compared with 47 workers not exposed to *n*-butanol, but ex-

posed to noise levels that ranged from 90 to 110 dBA. Mean age and exposure duration were not different between the groups.

Trichloroethylene

Another study demonstrated bilateral, high-frequency sensorineural hearing loss in 26 out of 40 workers exposed to excessive concentrations (above international recommended values at that time) of trichloroethylene (TCE).²⁸ Cases with previous or current noise exposure were excluded from the study. When the health status of populations ex-

posed to TCE through contaminated water ($n = 4281$) were investigated in the TCE Subregistry of the National Exposure Registry, a significant increase in reported hearing losses was found for the 0–9 years-of-age group.²⁹

Carbon Disulfide

Three studies were conducted with workers from viscose rayon plants, where exposure to carbon disulfide constitutes its main hazard.^{19,30,31} The first two studies compared workers with simultaneous exposure to excessive levels of carbon disulfide (that ranged from 2 to 10 times international recommended limits) and noise (86–89 dBA) to workers exposed exclusively to the same noise levels. An increased prevalence of high-frequency sensorineural hearing loss was observed in both investigations.^{19,30} Groups of workers with various exposure histories to carbon disulfide were compared to nonexposed workers.³¹ ABRs were significantly altered only for the group with the longest exposure (more than 20 years) and a history of excessive exposure. The results suggested that chronic exposure to carbon disulfide in humans affects the ascending auditory tract in the brainstem.³¹

n-Hexane

Studies that investigated workers exposed chronically to excessive concentrations of *n*-hexane reported abnormal ABR results that indicated slow neural conduction times.^{32,33}

Styrene

Workers exposed to low levels of styrene did not appear to have increased age-dependent hearing loss at high frequencies.³⁴ However, a comparison within the group of exposed workers between the least exposed and the most exposed revealed a statistically significant difference in hearing thresholds at high frequencies. Routine hearing tests of workers exposed to styrene in a plastic boat plant did not indicate hearing losses resulting from causes other than exposure to noise.³⁵ Nevertheless, 7 of 18 workers displayed abnormal results in central auditory system testing. Sty-

rene and noise exposures were meticulously assessed for 299 workers in the fiber reinforcement industry.³⁶ Noise levels were found to be in the range between 85 and 90 dBA, while styrene levels were generally below the recommended level of 50 ppm. The association between noise exposure, based on the developed lifetime noise dose estimate, and hearing loss was significant. That was not the case for styrene exposure. Styrene exposure approached significance for hearing loss only at some specific frequencies.³⁶ The earlier studies on styrene did not analyze hearing status as a binary variable.^{34,35}

Toluene

A study of 190 workers was carried out with rotogravure printing workers.³⁷ The hearing and balance functions of a group of printers exposed simultaneously to noise (88–98 dBA) and toluene (100–365 ppm) were compared with a group of printers exposed to noise alone (88–97 dBA), a group exposed to a solvent mixture in which toluene was the major component, and a group neither exposed to noise nor toluene. The adjusted relative risk estimates for hearing loss were 4 times greater for the noise group; 11 times greater for the noise and toluene group; and 5 times greater for the solvents group. Acoustic reflex measurements suggested that the hearing losses found in the group exposed to both agents might be due, in part, to lesions in the central auditory system.

The effects of toluene on the auditory system were studied in a group of rotogravure printers through the use of ABR.³⁸ Forty workers with normal hearing ability (assessed by pure-tone audiometry), who had been exposed to an average of 97 ppm for 12–14 years, were selected to participate. Their results were compared with those from a group of workers of the same age but not occupationally exposed to solvents. The study indicated that exposure to toluene induced a statistically significant alteration in the evoked responses, visible for all waves and all the intervals studied. The ABRs demonstrated auditory nervous system modifications before

the occurrence of neurological clinical signs due to chronic exposure to toluene.

Solvent Mixtures

Studies were conducted on auditory and vestibular functions of workers exposed to a mixture of unspecified alcohols, jet fuels, and aromatic solvents.^{39,40} The findings of pure-tone audiometry and speech discrimination testing were essentially normal for age and noise exposure history, not indicating measurable cochlear damage to solvent exposure. However a significant abnormality was found in tests that assessed more central portions of the auditory pathways.^{39,40}

In a study that investigated the effects of combined exposure to noise and a mixture of solvents that included toluene, benzene, styrene, xylene, and butyl acetate, an increased prevalence of hearing disorders was found.⁴¹ Almost half of the workers reported hearing loss, which was documented by audiometric testing to be permanent hearing losses of as little as 10 to as much as 60 dB HL.

Workers exposed to a mixture of polystyrene resin, methanol, and methyl acetate at levels below limit values were evaluated for auditory sensitivity.⁴² Findings were analyzed as the percentage of subjects falling below the 90th percentile of the upper limit of hearing. The percentages were 8.7 for the control group, 12.1 for the noise-exposed group, and 33.3 for the solvents and noise-exposed group. In the combined exposure group, styrene was the only solvent likely to be ototoxic itself. However, the combination of any of the several solvents, or noise and solvents, could have played an important role in causing this effect.

The relationship between self-assessed hearing disorders and occupational exposure to solvent mixtures was investigated in a cross-sectional design with 3284 men.¹⁰ Exposure to solvents for 5 years or more resulted in an adjusted relative risk for hearing impairment of 1.4 in men without occupational exposure to noise. A subsample of 51 men was examined with pure-tone audiometry and 20 of the 21 men who reported abnormal hearing

also fulfilled an audiometric criterion for hearing impairment. Occupational exposure to noise had an effect twice that of solvents; and in the case of combined exposures, the effects from noise dominated.

Metals

Table 35-2 displays the experiments on the effect of exposure to metals on hearing sensitivity in humans.

Lead

Workers exposed to lead acetate have been reported to experience sensorineural hearing loss.⁴³ Children considered to be at risk for lead intoxication from peeling lead paint or dust brought home from parents had their auditory systems examined through ABR.⁴⁴ The results of children with elevated blood levels showed that the latencies of Waves III and IV increased linearly with blood lead level, indicating a slowing of auditory nerve conduction velocity due to lead exposure. A 5 year follow-up of the children with low to moderate lead exposures showed persistence of the prolonged latencies at repeat testing.⁴⁴ Audiometric data were obtained by the Second National Health and Nutrition Examination Survey on 5717 children, most of whom had blood levels measured.⁴⁵ The probability of elevated hearing thresholds increased significantly with increasing blood lead levels. ABR^{46,47} and auditory event-related potentials⁴⁸ were recorded in workers exposed occupationally to lead who had their blood lead levels monitored. Blood lead levels were significantly correlated with abnormalities in the recorded evoked potentials.⁴⁶⁻⁴⁸

Mercury

Although less pronounced than the alterations caused by lead exposure, significant alterations in the ABR were also observed in the case of occupational exposure to mercury.⁴⁷ Mercury intoxication has been associated with hearing loss.⁴⁹ Up to 80% of patients treated for the fatal Minimata disease (due to ingestion of mercury contaminated food

Table 35-2 References on Effects of Exposure to Metals on Human Auditory System

Toxicant	Effect	Reference
Lead	Sensorineural hearing loss and vertigo Prolonged Waves III and V latencies in children with elevated blood lead levels Probability of elevated hearing thresholds increased as blood levels increased	Ciurlo and Ottoboni ⁴³ Otto et al. ⁴⁴ Schwartz and Otto ⁴⁵ Discalzi et al. ^{46,47}
Mercury	Central auditory disorders High prevalence of retrocochlear hearing loss (80%) in cases of intoxication Prolonged I–V interpeak latency for mercury exposed	Araki et al. ⁴⁸ Kurland et al. ⁴⁹ Mizukoshi et al. ^{50,51} Discalzi et al. ⁴⁷
Methylmercury	Mild hearing loss that extends over almost the entire frequency range to total deafness	Wustner et al. ⁵³ Gerstner and Huff ⁵⁴ Amin-Zaki et al. ⁵²
Manganese	Prevalence of hearing loss higher for manganese-only exposed than for non-exposed workers; noise exacerbated and accelerated manganese-induced loss; vestibular disorders	Nikolov ⁵⁵
Arsenic	Prevalence of low-frequency hearing loss higher for arsenic-exposed than for nonexposed children	Bencko et al. ⁵⁶ Bencko and Symon ⁵⁷

and water) suffered hearing loss. Long-term follow-up studies have been reported.^{50,51} Twenty-eight percent of the patients retested showed deterioration of hearing; 7% showed an improvement. Békésy audiometry and the Short Increment Sensitivity Index indicated that the early and middle stages of mercury intoxication may have resulted from cochlear lesions, whereas hearing impairments in late stages may have resulted from retrocochlear lesions.⁵⁰ Brain autopsy studies of the mercury intoxicated patients showed demyelination in the temporal lobes and heavy deposition of heavy metals in the transverse temporal gyri.⁵¹

Methylmercury

Twenty-four out of forty-nine children who ingested seed grain contaminated with methylmercury suffered from hearing disorders that ranged from elevated auditory thresholds to total deafness.⁵² A case of hearing loss was reported following the use of a mercury-based

product to remove freckles.⁵³ Hearing impairment is reported to develop early in the case of methylmercury poisoning. The hearing loss extends over almost the entire frequency range. In the most serious cases, the hearing loss can be profound.⁵⁴

Manganese

Altered hearing and balance functions have been reported in a study that examined workers exposed to manganese alone, or in concert with noise.⁵⁵ Pure-tone audiograms of manganese-exposed workers were affected in both low and high frequencies. Manganese ototoxicity appeared to be accelerated and exacerbated by noise exposure.

Arsenic

An epidemiologic study was conducted on a population of children living by a plant responsible for an emission of arsenic in the air.^{56,57} Analysis of the children's hair, blood, and urine revealed arsenic content. Signifi-

cant hearing losses were observed in the low frequencies of the audiogram when they were compared with nonexposed children.

Asphyxiants

Carbon Monoxide

Cases of accidental carbon monoxide poisoning that caused severe neurologic and psychiatric symptoms and hearing impairments that partially improved with time were reported.^{58,59} A 78% prevalence of sensorineural hearing loss among 700 cases of carbon monoxide intoxication was observed.⁶⁰ ABRs were studied in 32 patients with acute carbon monoxide poisoning.⁶¹ The abnormalities observed were divided into two patterns: a peripheral pattern of prolongation of Wave I latency without the prolongation of interpeak latency (six cases), and a central pattern of prolongation of latencies for all waves and interpeak latencies (two cases). The prevalence of ABR abnormality increased with the duration of unconsciousness.⁶¹

Discussion

In most cases, the main objective of the investigations reviewed was to examine either the neurologic effects of the chemical exposure or the general toxicity of chemicals. In the case of solvents however (Table 35-1), the majority of the investigations specifically focused on auditory system disorders. The multiplicity of objectives is reflected in the variety of study designs that include case reports as well as case-referent and cross-sectional studies.

A wide variety of metrics have been utilized, including pure-tone audiometry, reflexometry, reflex decay, speech audiometry, auditory-evoked potentials (ABR, cortical response audiometry, auditory event-related potential-P300), Short Increment Sensitivity Index, Békésy audiometry, clinical balance screening, electronystagmography, brain autopsy, and self-assessment.

Diverse approaches were used in the evaluation of chemical exposures. An appropriate assessment of chronic chemical exposures represents one of the major challenges in tox-

icologic investigations. In most of the reported investigations, the exposure history was evaluated through questionnaires combined with sparse exposure records. Unfortunately, this was also the case regarding noise exposures that certainly may constitute the main confounding factor in these investigations. In some cases, noise exposure was not even taken into account.

It is common in review studies such as this for the authors to raise new questions instead of providing answers to old ones. That is exactly what we have done here. The studies reviewed are hardly directly comparable, but nevertheless they provide indications that environmental and industrial chemicals may affect human hearing and may interact with noise, thus further exacerbating the situation.

There is indication that some of the chemicals reviewed affect auditory functions in animals through different modes of exposure.^{15,20,21} Extrapolation of the results from animal studies to humans should be made with caution. The frequency range of hearing and the metabolism of chemicals are different between animals and humans. In addition, the chemical concentrations used in animal experiments are higher than common occupational levels in Europe and the United States. Solvent abusers are, however, often exposed to similar or even higher concentrations^{62,63} and some occupational exposures of the same order of magnitude have been measured as well.^{19,30,31,37} Moreover, it is still common nowadays for high peak exposures to occur in the work environment due to the misuse of solvents to clean machinery, clothing, or hands, to mop floors, etc.⁶⁴

In animals, there is evidence for a cochlear site of the damage caused by certain solvents indicating that ototoxicity is most likely the process by which they affect the auditory system.^{9,18,20,21} However, reports on occupational exposures have indicated that the observed disorders had central auditory pathways involvement, suggesting a neurotoxic action of the solvent, probably in conjunction with its ototoxicity.^{31,35,37,40} If such is the case, hearing tests such as ABR and acoustic reflex measurements could contribute not only as

early indicators of those who are more susceptible to the development of hearing loss, but may also help to identify those who are more susceptible to eventual neurotoxic effects of solvent exposure or an unfavorable interaction between agents.

A report on a geriatric population ($n = 871$) with subgroups of workers with a history of industrial chemical exposures failed to demonstrate an interaction.⁶⁵ Conversely, an interaction between noise and chemical exposure was indicated by the finding that the hearing loss in workers exposed to both agents was greater than that expected from the noise exposure alone.^{10,13,14,36,42,55,66} The fact that detailed noise and chemical exposure records were not always available, together with the limited information that pure-tone audiometry can offer, constitute the main difficulty in reaching a conclusion from these reports as to hazardous exposure levels.

Nevertheless, the implications of the assumption of a possible auditory effect of chemicals are profound. Because there is evidence that exposure to a chemical alone, or in combination with noise can produce hearing loss, it is likely that current hearing conservation practices are not meeting the needs of populations exposed to chemicals. The evidence presented here should encourage further investigations in the following topics: mechanisms of lesion, dose-response estimations, the adequacy of pure-tone audiometry for monitoring solvent-exposed workers, the appropriateness of current hearing conservation practices, and finally, the role of hearing assessment as applied to the early identification of those most susceptible to neurotoxic disorders. The understanding of the effects of combined exposures will allow more comprehensive and adequate planning of hearing loss prevention strategies.

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Chapter 36

Review of Nonauditory Effects of Blast Overpressure

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A basic understanding of the mechanics of blast injury was developed during and just after World War II.¹⁻¹² Research since that time has further added to the understanding of the effects of classical or Friedlander blast waves.¹³⁻¹⁷ One clear finding is that the gas-containing organs are more vulnerable to direct blast than are the solid organs.¹⁸⁻²¹ This has led to the development of damage-risk criteria for blast injury in humans.^{22,23} These criteria assume the lungs to be the most vulnerable organ in terms of immediate pathophysiological effects. Disruption of the lungs by a blast wave can lead to upper airway obstruction from hemorrhage, deleterious changes in blood-gas parameters from increased venous/arterial shunting, and congestive heart failure from increased pulmonary hypertension. At higher overpressure levels, the formation of alveolar-venous fistulae in the lung parenchyma permits air to enter the circulation. This can lead to early death from coronary and cerebral air embolism. Contusions and/or ruptures of the lining of the gastroenteric tract are also important. In addition, the onset of the effects of these lesions become more important with time. Criteria have also been established relating the severity of eardrum rupture to blast overpressure and duration.²⁴

Recent studies have shown that the upper respiratory tract can be important as an indicator of the possible presence of other nonauditory blast injuries during the initial medical assessment.²⁵⁻²⁷ It was observed that demon-

strable hemorrhagic changes tend to occur in the upper respiratory tract either before or concurrently with injuries to the gastroenteric tract and/or the lungs.

Data from the effects of complex wave experiments have shown that much of the results from free-field waves may be applied to complex waves.^{28,29} However, complex waves may take on many forms, not all of which data are available to properly assess the risk. But for the specific case in which firing a rocket launcher from an enclosure is simulated, work has just been completed that establishes some noninjury thresholds.³⁰

Some Physical Characteristics of Blast Waves

A brief discussion of some of the basic physical characteristics of both classical and complex blast waves is essential to the understanding of the nonauditory effects of the various waveforms.

Friedlander Waves

The classical or Friedlander blast wave occurs in an open, free-field environment. It is characterized by nearly an instantaneous rise in pressure to some peak value followed by an exponential decay to some negative value before returning to ambient pressure (Figure 36-1).³¹ The overpressure phase duration is shorter than the underpressure phase.

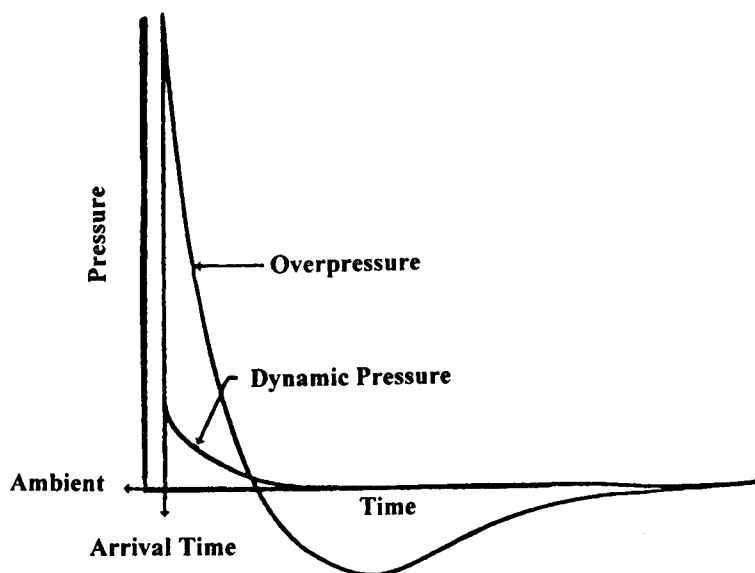


Figure 36-1 Characterization of a classical Friedlander blast wave with associated dynamic pressure component.

Complex Waves

Complex waves can be described as a series of nonuniform blast waves separated by varying time intervals. They can be produced in many ways, some of which are: blast interaction with obstacles, entry of a blast into an enclosure, firing a rocket launcher from an enclosure, and explosions in an enclosure.

The pressure–time history illustrated in Figure 36-2 was recorded during a Carl–Gustaf recoilless weapon simulation in a vented $3.05 \times 2.44 \times 2.44$ m enclosure using an instrumentation cylinder.³⁰ Note the variability of the waveform.

Bioeffects of Blast

Damage-risk criteria for primary blast effects from classical Friedlander waves were established in 1968 and have often been called the Bowen curves.²² Attempts to use the same criteria for a complex wave environment are complicated by the variability of the dynamic components of the waves and an all-inclusive criterion is not yet possible.

Friedlander Waves

Numerous mammalian mortality studies have demonstrated that tolerance to classical blast waves is dependent upon animal species and the peak overpressure and overpressure phase duration of the blast wave.^{16,32,33} Analyses³⁴ of the mortality data for animals exposed against a reflecting surface illustrate three important points (Figures 36-3a and b).¹⁶ First, for a given species and wave duration, there is a linear relationship between the probit of mortality and the logarithm of reflected overpressure; second, all the lines have a common slope that indicates the same standard deviation and suggests a common mechanism of lethality; and third, the data tends to separate into “small” and “large” mammal groups. The probit transformation used in these and the forthcoming analyses is simply the addition of five standard deviations to the normal distribution of the deviate to eliminate negative values. Thus, a probit value of 5 is equal to the median or the 50% mortality.

By combining scaling laws derived from dimensional analyses with additional probit an-

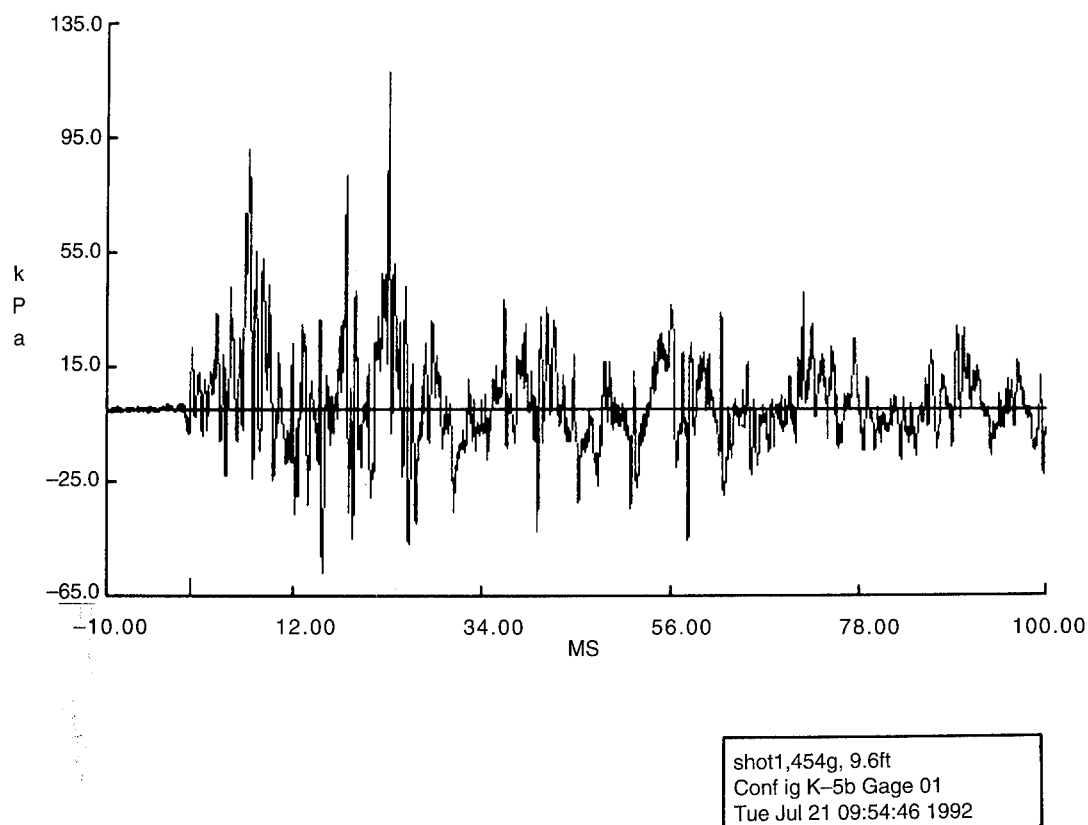


Figure 36-2 Pressure-time recordings for gauge 1 on an instrumentation cylinder during a Carl-Gustaf recoilless weapon simulation.

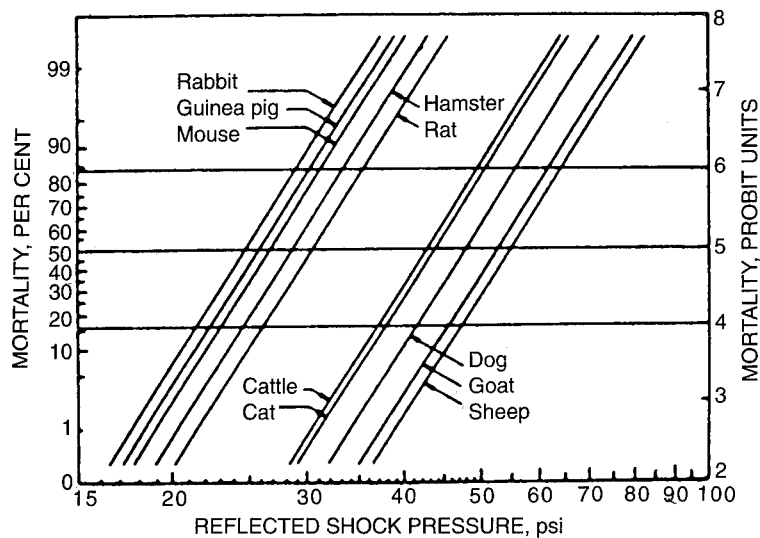
analyses of the mortality data used to derive Figures 36-3a and b, the experimental dose to a mammal can be used to calculate the equivalent overpressure, P^* (psi),²² that would produce the same mortality in a 70 kg mammal exposed to a long duration blast wave (i.e., square wave) at sea level:

$$P^* = P / (1 + 6.76T^{-1.064}) \quad (36-1)$$

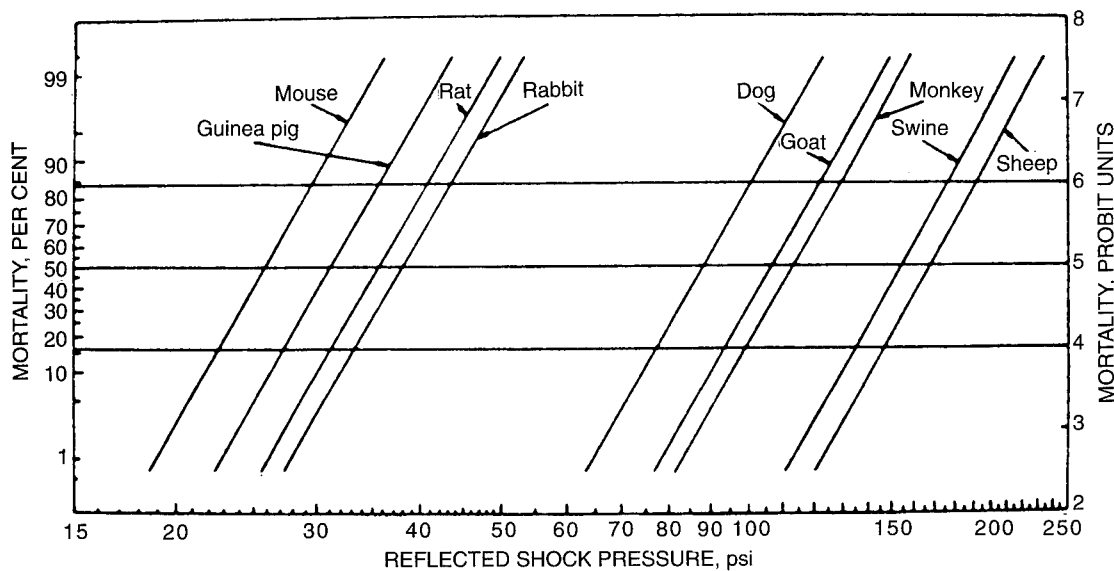
where $P = P_r(14.7/P_o)$ and $T = t_+(70/m)^{1/3}$ ($P_o/14.7$)^{1/2}; P_r (psi) is the experimental peak reflected pressure; and P_o (psi) is the ambient pressure. The duration, T (milliseconds), is derived from the experimental duration, t_+ (milliseconds), and m (kg) is the body mass. Using the equivalent overpressures, both individual and parallel probit analyses were done for all the species listed in Figures 36-3a and b, except that the guinea pig was ex-

cluded from the parallel analyses. Figure 36-4 is an illustration of the sheep mortality data treated in this manner.²² Note that the agreement is much better for the median than at the extremes of the distribution and that parallel analysis narrows the confidence bands at the extremes. Nevertheless, the reader should bear in mind that the agreement is not as good at the 1% mortality or 99% survival levels. Projecting this level to a no-injury level must be done with caution.

The 13 calculated square-wave overpressures, P_{sw} 's, listed in the table in Figure 36-5 can be considered as indices of blast tolerance that are independent of body mass.²² The p 's for the eight large animal species ranged from 50.0 to 71.9 psi with a geometric mean of 61.5 psi, and those for the five small animal species ranged from 30.8 to 36.9 psi with a geometric



(a)



(b)

Figure 36-3 (a) Mortality curves for animals exposed to long duration reflected pressures while mounted side-on against the end plate of a shock tube. Probit regression equation: $y = a + b \log x$; where y is the percent of mortality in probit units, a and b are the intercept and slope constants, and x is the pressure. All measurements were made at an ambient pressure of 12 psia.¹⁶ (b) Mortality curves for animals exposed to short duration reflected pressure from high explosive charges detonated overhead while mounted prone on a concrete pad. Probit regression equation: $y = a + b \log x$; where y is the percent of mortality in probit units, and a and b are the intercept and slope constants, and x is the pressure. All measurements were made at an ambient pressure, 12 psia.¹⁶

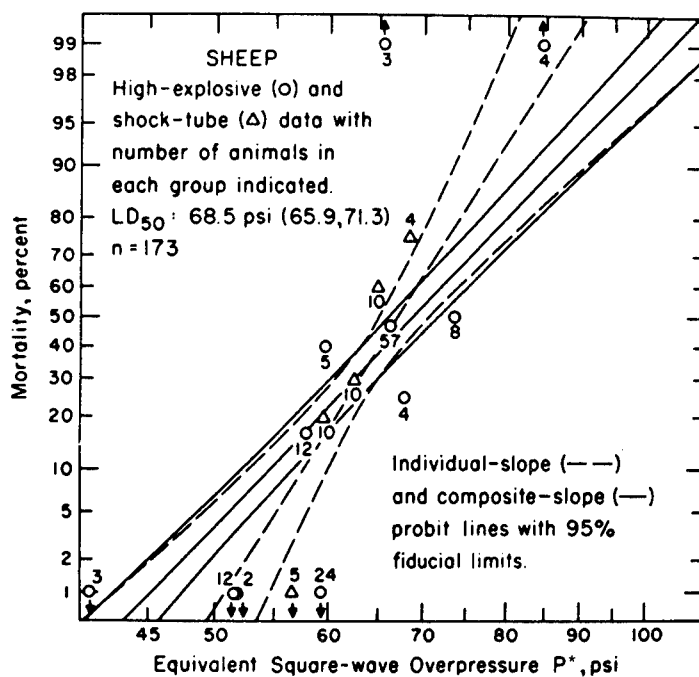


Figure 36-4 Results of the parallel-probit analysis for the sheep in terms of equivalent square-wave overpressure, defined by $P^* = P/(1 + 6.67T^{-1.064})$. Results of an individual analysis for the sheep are shown for comparison.²²

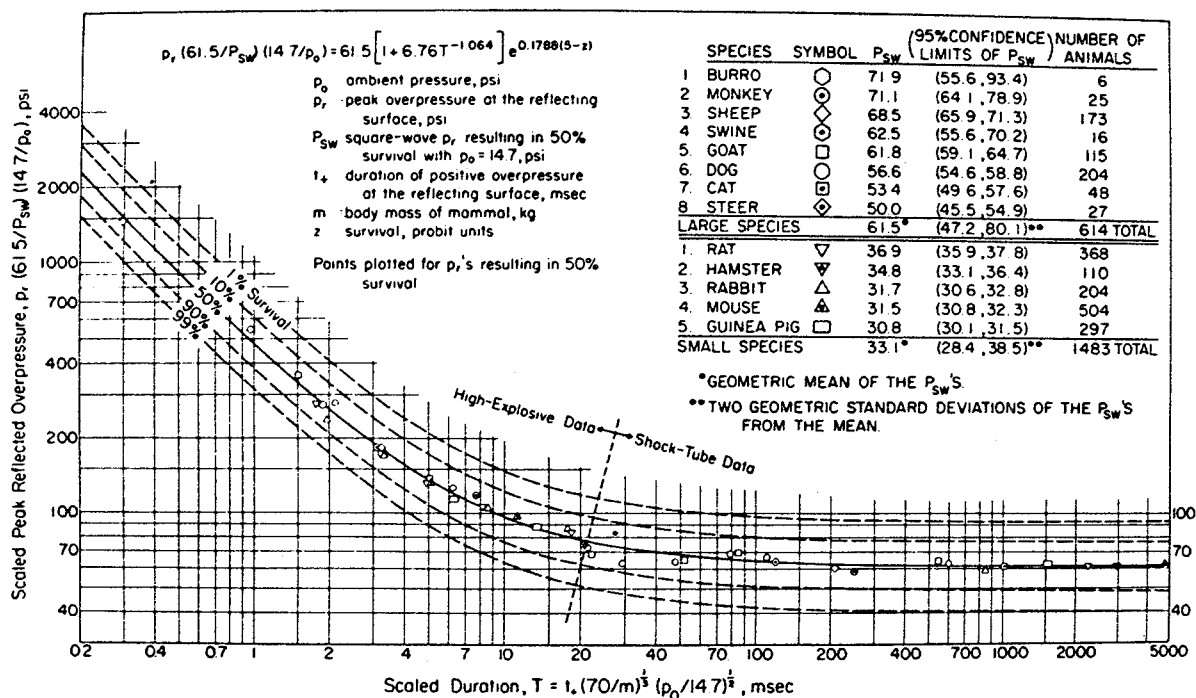


Figure 36-5 Twenty-four hour survival curves, applicable to sharp-rising blast waves, derived from the analysis of data for 12 mammalian species (excluding guinea pig).²²

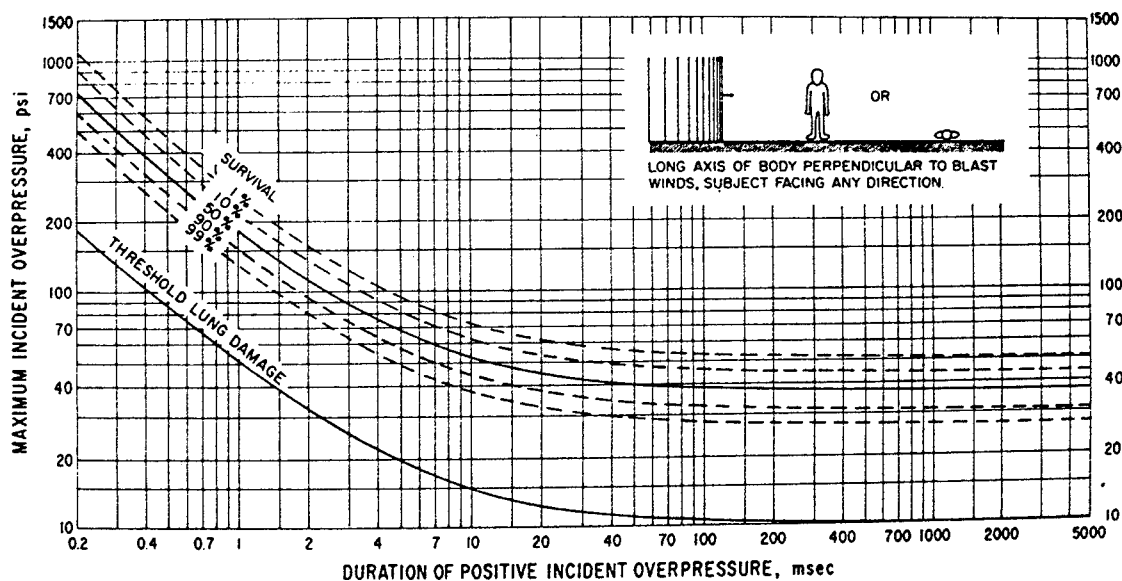


Figure 36-6 Survival curves predicted for 70 kg man applicable to free-stream situation where the long axis of the body is perpendicular to the direction of the propagation of the blast wave.²²

mean of 33.1 psi. This large/small breakdown is supported by the fact that the ratio of lung volume to body mass is distinctly different between large and small animals. Lung volume and density data, as well as whole-body impact tolerances, clearly suggest that humans belong to the large animal group.^{9,16,33,35,36} Assuming this to be true, the tolerance for a human was arbitrarily taken to be the geometric mean of this group ($P_{sw} = 61.5$ psi).²²

Survival curves giving peak reflected overpressure versus positive overpressure duration scaled to an ambient pressure of 14.7 psi and a 70 kg mammal with a P_{sw} of 61.5 psi, can be derived from the equation:

$$P = 61.5[1 + 6.76T^{-1.064}]e^{0.1788(5-z)}. \quad (36-2)$$

The equivalent overpressure was made applicable to various levels of lethality by the transformation

$$P^* = P_{sw} e^{c(y-5)} \quad (36-3)$$

where y is the mortality in probit units and c is the reciprocal of the probit slope (which is 0.1788 in this case). Because the curves are presented in terms of survival, the term $e^{c(y-5)}$

becomes $e^{c(5-z)}$, where z is the survival probit. The overpressure scaling factor, $61.5/P_{sw}$, seen in Eq (36-2) accounts for differences in species tolerances. This technique allows all points to be compared to the 50% survival curve derived for humans.²²

The human damage-risk curves for a standing person, illustrated in Figure 36-6, were derived directly from this set of curves. There are also curves for prone individuals as well as those standing next to a reflective surface; the longer the duration, the lower the pressure required to produce injury. The importance of duration declines and at durations longer than 50 milliseconds, the peak pressure is nearly constant. The peak pressure necessary for trauma induction is a function of the orientation of the individual with respect to the blast. Subjects in a prone position, with the long axis of the body parallel to the blast wind, are more resistant to injury than those that are standing or prone broadside with the long axis of the body perpendicular to the blast wind for the same incident blast wave. If the long axis of the body is parallel to the blast wind, the incident pressure is the blast load; but if the axis is parallel to the shock front, the stagnation

pressure (i.e., incident plus the dynamic pressure) is the load. Individuals near a reflecting surface that is parallel to the shock front are the most vulnerable for the same incident overpressure if the positive phase duration of the wave is spatially longer than the width of the body.²² In this case, the incident wave upon striking the reflecting surface produces a reflected wave with a peak overpressure more than twice the original value.³¹ This reflected pressure is then the load on the body.

However, if the positive phase duration is spatially shorter than body width, the decay rate of the blast wave is such that the most severe loading occurs on the side of the body closest to the oncoming wave. The impulse of the blast wave at the leading edge of the body becomes the effective dose and not the reflected pressure.²²

Complex Waves

The response of mammals to complex blast waves is more difficult to interpret. Subjects may be more or less tolerant to complex waveforms than to classical waves. The rate of rise, the number and intensity, and the frequency of oscillation of the pressure pulses are among the additional parameters that are important in determining biological tolerance to complex waves.

To illustrate the complexity of complex waves it is useful to mention some of the better known incidents and studies. An incident occurred during World War II in which 13 men occupying an antiaircraft gun emplacement were exposed to the detonation of a 2,000 lb HE bomb.^{10,37} The bomb exploded 9.2 m from the 4 × 6 m emplacement that was open at the top. It was surrounded by a 1.6 m high rampart with a single entryway. Two men crouched against a wall in the corner farthest from the blast were fatally injured, whereas two subjects closer to the blast sustained only slight injuries. The two individuals that had slight injuries and no eardrum ruptures were near the upstream wall. They were probably in a relatively low pressure region created by the vortex that was formed as the blast wave swept over the rampart.

Wave Entry Into Enclosures

During the above-ground nuclear weapons test period, several species of animals were exposed to the complex waves formed in the fast and slow fill chambers of underground shelters.^{37,38} Except for eardrum rupture, there was no direct correlation between primary blast injuries and the maximum overpressure and the duration of the pressure-time patterns recorded in the shelters. The incident and reflected pressure spikes associated with the diffraction phase were either not large enough individually or the time intervals between shocks were long enough so the rates of pressure rise in the gas-containing organs were too slow to produce significant levels of nonauditory injuries.³⁹ The high-velocity flow through the entryway of the fast-fill chamber posed the most significant hazard to the test subjects. On two separate field studies, the dogs that were placed near the entryway of the fast-fill chamber were either killed or sustained serious injuries from violent displacement and subsequent impact with the rear wall of the chamber.

Blast jetting through open hatches and firing ports, and not the complex wave formed after, is also the major direct airblast hazard to personnel inside an armored vehicle for long duration blasts. Studies with sheep placed in front of the openings indicated that slight crushing type injuries began at jet flow stagnation levels above 10.3 psi and that severe injuries would be predicted at stagnation levels over 20.6 psi. These criteria were limited to jetting through either circular openings 15–48 cm in diameter or equivalent area rectangular openings having length/width ratios of 1.0:2.5.⁴⁰

In structures with small volume to opening area ratios, such as foxholes, the direct overpressure effects of the diffraction phase predominate because of the multiplication of the incident shock front reflections from the walls and floor of the foxhole. There is little or no fill phase associated with these open structures. The magnitude of the reflected shocks that represent the peak overpressure can be more than two times the pressure in the incident

wave. The results of laboratory studies with rats in one-seventh scale model foxholes and on field tests with sheep in full-scale two-man rectangular foxholes demonstrated that response tended to vary with the time interval between incident and reflected shock for a given incident overpressure.^{41–43} These findings correlated quite well with the results from steploading experiments in which subjects were exposed to reflected shocks at various distances from the end plate of a shock tube.^{43–45} As the distance from the end plate and correspondingly the time between the incident and reflected shocks increased, the lethal peak reflected overpressure needed rose to about 1.8 times that from a long duration Friedlander wave. The critical time delay for increased resistance varied with species size. It was approximately 0.13 millisecond for rats and 1 millisecond for sheep.

The protective effects of long duration pressure loading was also demonstrated by pressurizing animals to increasingly larger ambient pressure levels prior to blast exposure.⁴⁶ It was found that resistance to blast injury increased as the ambient pressure increased.

Firing From an Enclosure

The results of a study conducted in 1976 suggested there was a significant risk of nonauditory injury associated with firing large caliber weapons from inside enclosures.^{47,48} Rabbits were exposed from one to three times at various locations inside a nearly closed room to reverberant pressure waves. The approximately 500 milliseconds duration waves were generated by firing the Carl–Gustaf recoilless weapon. The interval between shots for the animals that were exposed two or three times was 1 minute. Nearly 35% of the animals sustained moderate to severe injuries from peak overpressures that never exceeded 40 kPa (186 dB). This pressure level is about one-fifth that required to produce the same level of injury in rabbits exposed to classical blast waves.⁴⁹ Spectral analysis of the waveforms at the various locations suggests the more severely injured animals were exposed to waveforms with the strongest pressure components oc-

curing in the 150–500 Hz range. This range matches the natural frequency of the rabbit thoracic–abdominal system.^{15,50,51} The authors suggest that the first 50 milliseconds of exposure to such a waveform would be enough to stimulate resonance thereby enhancing injury.⁴⁷ However, the differences in injury levels could just as easily be attributed to the number of blasts the rabbits received. Eighty-four percent of the rabbits with lung weight/body weight hemorrhagic ratios of 1.50% or more were exposed to two or three blasts; whereas, only 41% of the rabbits with ratios of less than 1.50% were exposed two or three times.

The proposed resonance effect was further investigated at this laboratory. Sheep were exposed to blasts from two 3.63 kg charges. The intervals between blasts were varied from 0 to 14 milliseconds in 2 millisecond increments. Lung hemorrhage did not change significantly as a function of the time between blasts. Results from another series of tests suggested that lung hemorrhage might be more severe when the interval between blasts was 9.6 milliseconds and less severe when it was 3.7 milliseconds.

Explosions in an Enclosure

Direct blast effects produced by shaped-charge warheads and by detonating bare explosive charges inside enclosures were the subject of recent investigations at this laboratory. The LAW, DRAGON, and TOW warheads were fired through various thicknesses of armor into enclosures and 0.123, 0.25, 0.50, 1.0, and 3.0 lb charges were detonated in the centers of four different enclosure volumes of 194, 200, 300, and 640 ft³ each. Injuries sustained by the sheep placed at various locations in these volumes were evaluated in terms of existing damage-risk criteria and enclosure volume.^{22,23} Attempts were made to correlate the injuries produced with the mean overpressure predicted for a given charge weight and chamber volume,⁵² the experimental overpressure obtained by curve smoothing of the complex wave, or the summed partial impulse segments of the first few milliseconds of

the waveform. The results of these various methods were plotted against the damage-risk criteria isodamage curves. Even though no definitive conclusions could be drawn, the data collected suggested that:

1. of the damage correlates evaluated, the experimental overpressure appears to work the best in relating injury levels to the isodamage curves;
2. the frequency content of a complex blast wave is important in determining the extent of injury because animals at different ranges from a detonation in a given enclosure tend to sustain equivalent damage;
3. for a given charge weight detonated in the center of an enclosure, the response of a subject near a wall varies with the volume of the enclosure; and
4. subjects in a corner sustained more injury than a subject along a wall at the same distance.

Mathematical Modeling of Injury

A mathematical model of a two-chamber spring-mass system (two-lung model) was developed to predict the response of the thorax to blast waves.^{33,36,53} It is a single degree of freedom system in which chest wall response (displacement, velocity, and/or acceleration) and intrathoracic pressure can be calculated for different blast loads.

This model has since been simplified to a single chamber one-lung model that assumes that the blast load is acting simultaneously upon both lungs.⁵⁴ The mathematical equation for the model is nonlinear. Its response is dependent upon the amplitude and the frequency content of the pressure loading function. As the pressure goes up in the lung cavity, the lung becomes stiffer and its spring constant increases subsequently increasing the "resonance" frequency.

For example, for sinusoidal pulse loads of 50 and 200 kPa, the maximal intrathoracic pressures will be reached at 50 and 85 Hz.⁵⁴ For step waves of 50 and 200 kPa amplitude, the resonance frequencies are 107 and 206 Hz,

respectively. The corresponding maximal intrathoracic overpressure for these loads is 80 and 602 kPa with respective chest wall velocities of 2.6 and 9.1 m/s.

This one-lung model was used to illustrate the effects of complex blast waves inside an armored vehicle attacked by a shaped-charge warhead.⁵⁴ Intrathoracic pressures and chest wall velocities were twice as high in the back as in the driver's narrow space in the front of the vehicle. That difference was attributed to the linear dimensions of the inside of the vehicle resulting in different resonance frequencies in the blast wave (107 and 135 Hz) and consequently different thoracic responses to the complex blast waves. No large animals were tested in these experiments.

In Bowen's curve for threshold lung injury illustrated in Figure 36-6, individual points on the curve represent one and only one specific Friedlander blast wave with its corresponding peak overpressure and duration. Applying these pressure-time histories to the one-lung model showed that the maximum inward chest wall velocity varied from 3 to 4.5 m/s and the intrathoracic pressure varied from 20 to 150 kPa.⁵⁵ The lower numbers were for the shorter duration blast waves. As first suggested by Jonsson,¹⁷ we believe that velocity is a more definitive measure of injury than lung overpressure.

Sheep were exposed to very complex blast waves in various locations in 11, 18, and 36 m³ enclosures wherein high explosive charges were detonated.²⁹ The 18 m³ enclosure wave was also used to simulate the back blast from a Carl-Gustaf recoilless weapon.³⁰ Pressure-time histories were measured at the various locations using an instrumented cylinder with the approximate dimensions of the body of a sheep.

These measurements were applied to the one-lung model. As seen in Figure 36-7, it was found that sheep having a 50% incidence of small petechiae in the lungs had a calculated chest wall velocity of 3.5 m/s that corresponds to the predicted velocities for Bowen's threshold injury curve.

Note the fluctuation in velocity produced by a complex wave environment that is illus-

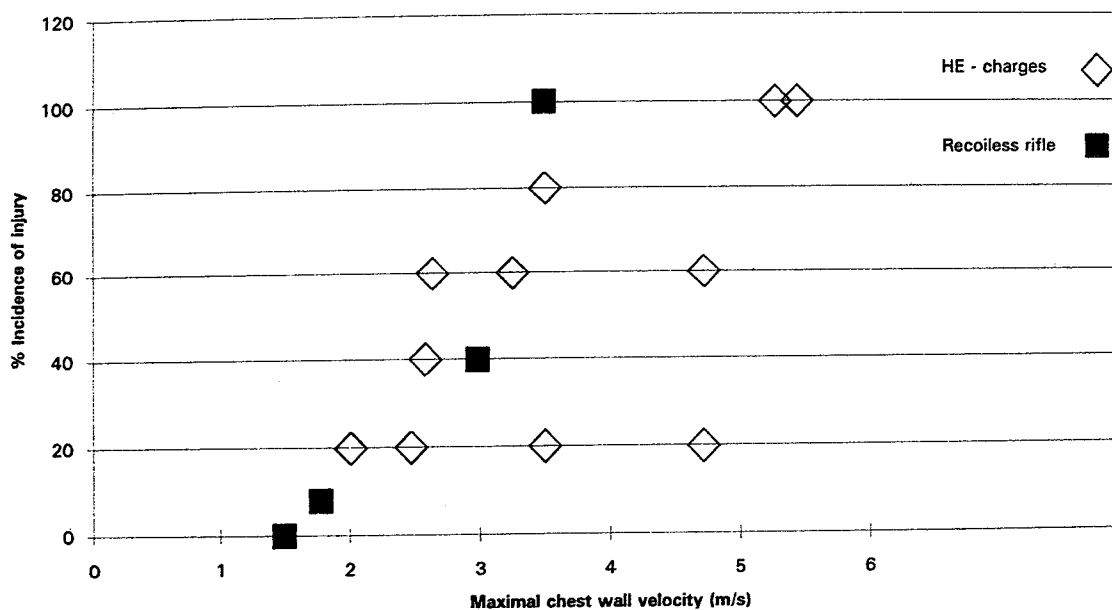


Figure 36-7 Percent incidence of lung petechiae versus calculated chest wall velocities using the lung model for the waveform of the recoilless weapon from bunker simulation model and input waveform described in Figure 36-8a and for HE-charge detonations in the bunker.

trated in Figures 36-8a and b. We believe that at exposure levels considerably above the threshold for injury, these multiple oscillations increase the injury. However, at threshold and subthreshold levels, we would expect that most oscillation frequencies would be safe.

Threshold of Injury

Aside from looking at a modeling approach, it is also useful to look at empirical data. The threshold lung injury curves from Eq (36-2) are presented in Figure 36-9. These curves were based on the incidence of small petechiae on the lungs of no more than 50% of the test animals. The free-field curve is for an individual standing away from objects with the long axis of the body perpendicular to the direction of propagation of the blast wave. The reflective curve is for persons next to walls or objects and generally differs from the free-field (free-stream) curve by at least a factor of 2. This factor accounts for the approximate doubling of the incident pressure wave due to the proximity to the flat surface. Because it is

not always possible to insure that personnel are away from reflective objects, the reflective curve is probably the better curve to rely on with respect to insuring against nonauditory effects. Although this curve may not be low enough for personnel in corners, such errors are only likely to result in minimal injury if these curves are indeed valid.

One of the bases for the curves illustrated in Figure 36-9 was data from dogs subjected to blast waves of 320 to 348 milliseconds duration.¹⁶ The point for 50% incidence of lung petechiae is plotted as the open circle. This point turns out to be a factor of 4 lower than the LD₅₀ curve. Therefore, the threshold curves assumed the shape of the LD₅₀ curves in which the incident pressure was reduced by a factor of 4. Further support of these curves came from a sheep study in which 64 lb high-explosive charges were used.¹⁶ While the lowest exposure level did show more than a 50% incidence of lung petechiae, it also plots above the reflected threshold wave.

Results of current studies conducted by the US Army also lend credibility to the validity of these curves. Because the US Army was con-

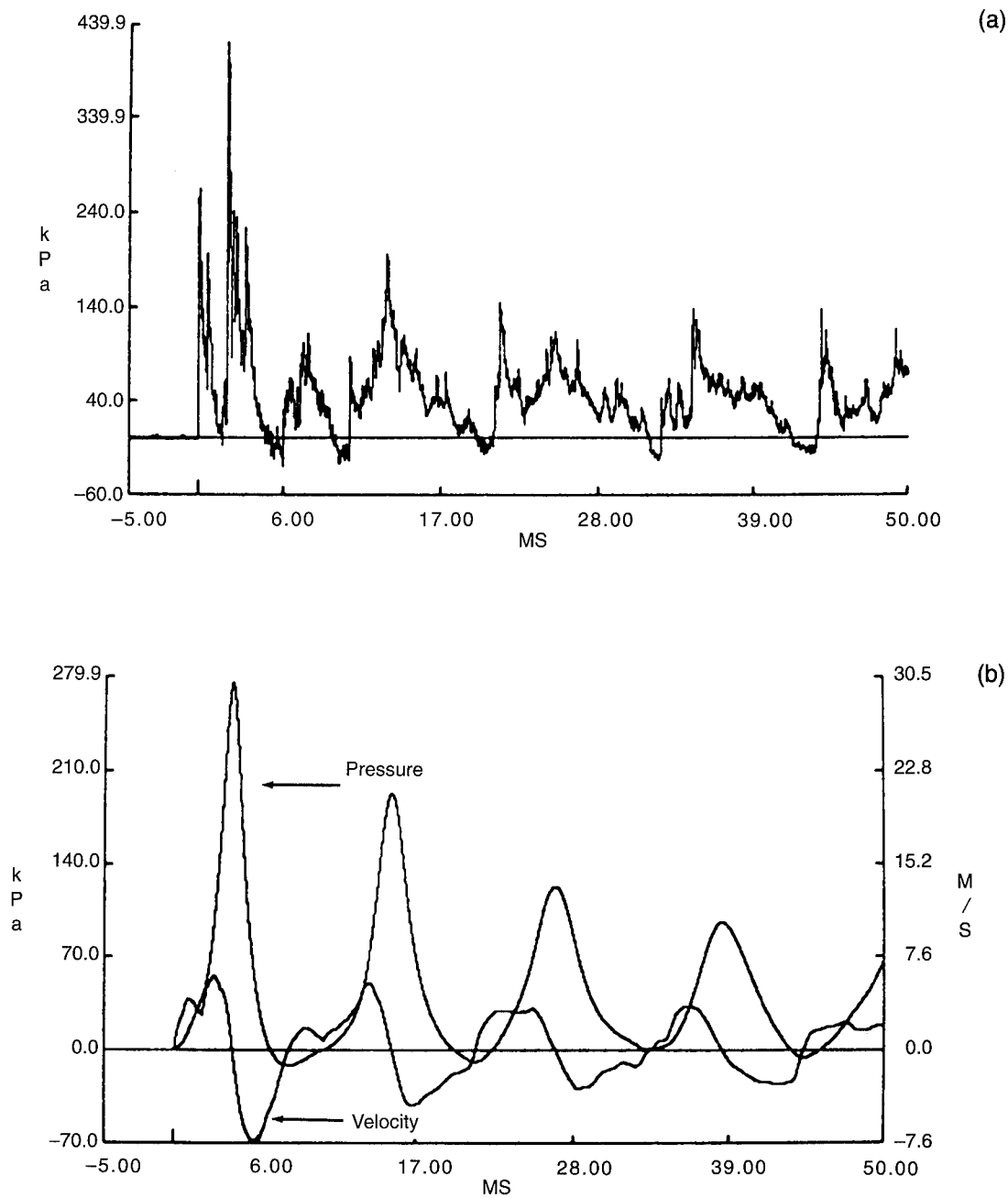


Figure 36-8 (a) Complex waveform from a charge fired in an enclosure. (b) The calculated chest velocities and internal chest pressures using the Bowen model and input waveform described in (a).

cerned about nonauditory risks from blasts during the training of soldiers, the Army began to use the Z curve seen in Figure 36-9 as a conservative nonauditory limit as well as a limit for hearing conservation while wearing

hearing protection. In fact, this Z curve is based on the US National Research Council Committee criterion for prevention of hearing loss from impulse noise.⁵⁶ Because the Z curve was considered likely to be very overprotec-

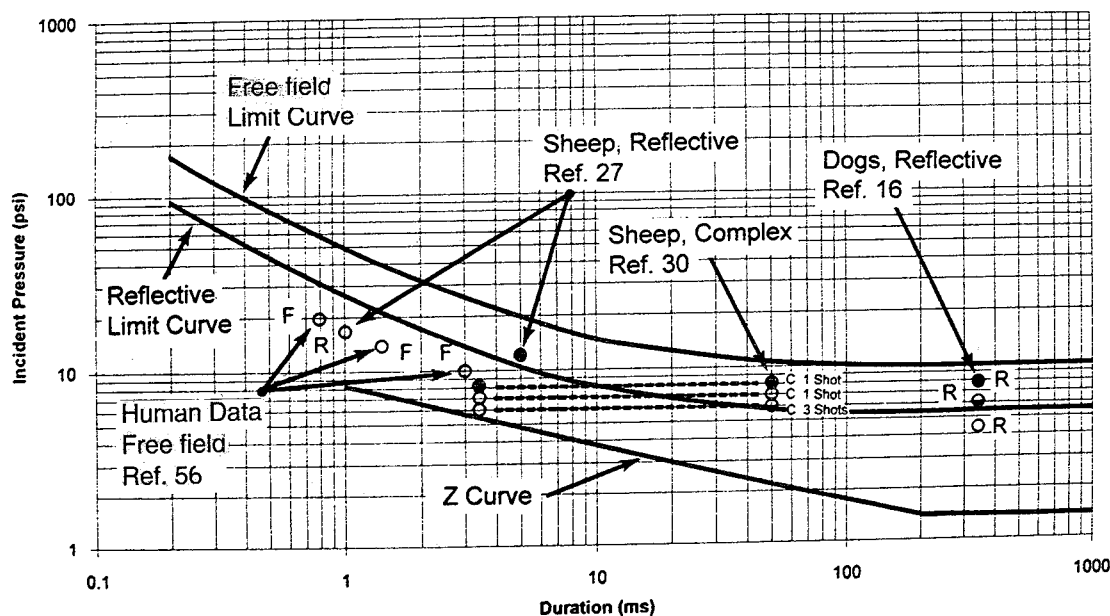


Figure 36-9 The threshold of lung injury from Eq (36-2) as well as the Z curve used in MIL-STD-1474C. Also plotted are data from references 16, 27, 30, and 56. The open circles were cases in which no lung petechiae were observed for the sheep or dogs. The half-filled circles indicate that one-half of the animals had some petechiae on the lungs. The solid circles indicate that some small isolated hemorrhages occurred. For the human studies, the lack of lung petechiae is assumed from the lack of petechiae on the larynx-pharynx. The F and R indicate the exposure was freefield or reflective, respectively.

tive with respect to nonauditory risk, some key studies linked to the blasts of actual and possible army weapons have been completed.

Current results of both human and animal studies are plotted in Figure 36-9. The human studies come from a final report of the blast overpressure studies recently finished at Kirkland AFB for the US Army.⁵⁷ At the highest peak pressures, which occurred six times at 1 minute intervals, with two exceptions, no nonauditory injury was observed. There were 104 subjects for the 190 dB, 3 milliseconds duration exposures; 58 subjects for the 193 dB, 1.4 milliseconds exposure; and 52 subjects for the 196 dB, 1.8 milliseconds exposures. One exception was a hematoma on the eardrum of one subject whose ear was protected by a leaking muff. The only other exception was a subject with bruised ribs (from playing football) who complained of great discomfort from the blast. Because the exposures fall just below the limit curves of Figure 36-9, the curves,

at least for these conditions, seem to be reasonable.

For complex waves, a set of threshold values for a firing from an enclosure simulation are available for sheep.³⁰ No-injury thresholds for one shot and three shots are located in Figure 36-9. The waveform of this simulation is shown in Figure 36-2. It is this set of data in which the Bowen model predicted a maximum velocity of 1.5 m/s for no injury.

The point of the dashed line between 3.5 and 50 milliseconds is that using classical techniques such as peak and duration, it is not clear how to assess a very complex wave as shown in Figure 36-2. If only the first part of the wave is used, the threshold of effect would be considerably lower than Bowen's reflective threshold. If the duration is taken as the B duration (the point in which the envelope decays to one-tenth of the highest peak), then the threshold of effect would plot above the Bowen's reflective curve. This case only illus-

trates some of the difficulty in assessing human response to a blast. Nevertheless, it is comforting to note that the dashed lines do cross the Bowen's threshold curves. The fact that there is an empirical difference between one-shot and three-shot thresholds for the firing from an enclosure simulation does point out that the number of exposures is a concern. Our attempt to try to blend the one-shot data of Eq (36-2) to the multishot data to obtain the safety limits of army training situations is forcing the issue. Yet, what we hope to have shown is that a logical pattern as to safety or threshold levels is slowly coming into place.

In summary, there exists a reasonable injury criteria for the blast effects for the free-field case. A reasonable extension of this criterion to one simple reflection also appears to be reliable. The use of the calculated chest wall velocity as predicted by a lung model also looks promising. However, there are many situations in which the exposure conditions are considerably more complex. For these cases, more research will be required to define the criteria for each case. The definition of safe limits for a firing from an enclosure simulation is a good first step. We believe that although the Z curve is probably too conservative, any exposure of personnel at exposure levels between the Z curve and reflective curve should be done with extreme caution. This is especially true for the exposure of personnel from their own weapons where it is possible for soldiers to increase the complexity of the exposure by their own actions.

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